Development of an Electrocardiographic Analysis Algorithm

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Abstract

There is a need to automate the present manual methods of analysing long term (24 hour) recorded Electrocardiogram (ECG) signals at high speed (60 times the recorded speed).

An algorithm has been developed which groups the recorded heart beats into one of up to six families depending on its shape and also into 'early/ on time/ late' depending on the time of occurance of a beat with respect to the preceding beats.

The algorithm was implemented on a general purpose mini-computer and supplied with data from the ECG tape replay unit. A total of 4608 beats so analysed were manually checked and found to be correct.

The construction of a stand-alone microprocessor implementation of the algorithm is proposed and possible extentions to the algorithm are discussed.

Acknowledgments.

I owe my thanks to many people who have helped me in various ways throughout the work presented in this thesis.

Firstly thanks go to my supervisor, Mr W. K. Kennedy for the long and patient hours he has spent ensuring that I stayed on the proper path, with discussions and advice. My thanks are also extended to Mr P. J. Bones from the Cardiology Department of Princess Margaret Hospital and to Dr A. E. McKinnon from the Christchurch Clinical School who have both been closely associated with the project from the start.

The staff at the Princess Margaret Hospital's Cardiology Unit also deserve my gratitude for answering my many questions in and around the subject of cardiology, and also for the use of their equipment. This project involved a number of computing hours and I would like to thank the Respiratory Department at Princess Margaret Hospital for the ready access to their computer.

I would also like to thank Miss A. R. Wilkinson whose artistic talents are displayed in the many illustrations in this thesis and whose comments made during the proof reading of the text have been invaluable.

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CHAPTER 1

Introduction.

A large body of information now exists which relates the clinical and pathological functioning of the heart to the various features of an electrocardiographic (ECG) waveform. Because an ECG signal is relatively easily obtained from a set of electrodes on the patients skin, and because valuable information on the condition of the patients heart can thus be obtained, the ECG is one of the more important diagnostic tools available to physicians.

ECG recordings are regularly made in doctors' surgeries and take from 5 to 15 minutes to perform. This enables the doctor to observe the patient's heart rhythm and will show if there is any deviation from the normal operation of the heart. However, because of the short measurement time, this method of ECG measurement does not necessarily show intermittent abnormalities that may be present.

As faults in the operation of a patient's heart are often random in nature, the current method of study is to record the patients ECG for 24 hours on a miniature tape recorder which is worn on a belt around the patients waist. The patient is then free to go about his

daily activities and yet any abnormalities that might occur can be analysed later by a trained technician.

It is impractical to replay the recorded tape in real time and the commonly used accelerated replay rate of 60 times real speed means a 24 hour tape can be scanned in 24 minutes, but this usually takes longer due to the accelerated replay being stopped to allow the real time replay of important events. This introduces another problem in that there are approximatly 100,000 heart beats on each 24 hour tape recording and even a trained and experienced operator is liable to miss important events during the analysis time, which is typically 3/4 to 1 hour, but which can extend to more than 3 hours.

What is therefore required is an automatic analysis system which summarises the recorded tape and allows for the rapid identification of possible regions of interest. This means that those periods of the recorded ECG which are of no immediate interest in a particular study can be bypassed, and the technician's time can be spent in a detailed manual analysis of only important recorded heart beats.

The system used to perform the analysis explained above is summarised in fig 1-1. The recorded tape is analysed by a computer which provides a summary of the patients heart activity over the recorded period.

This thesis presents the algorithm developed to do the computerised analysis. Its performance in a trial implementation is

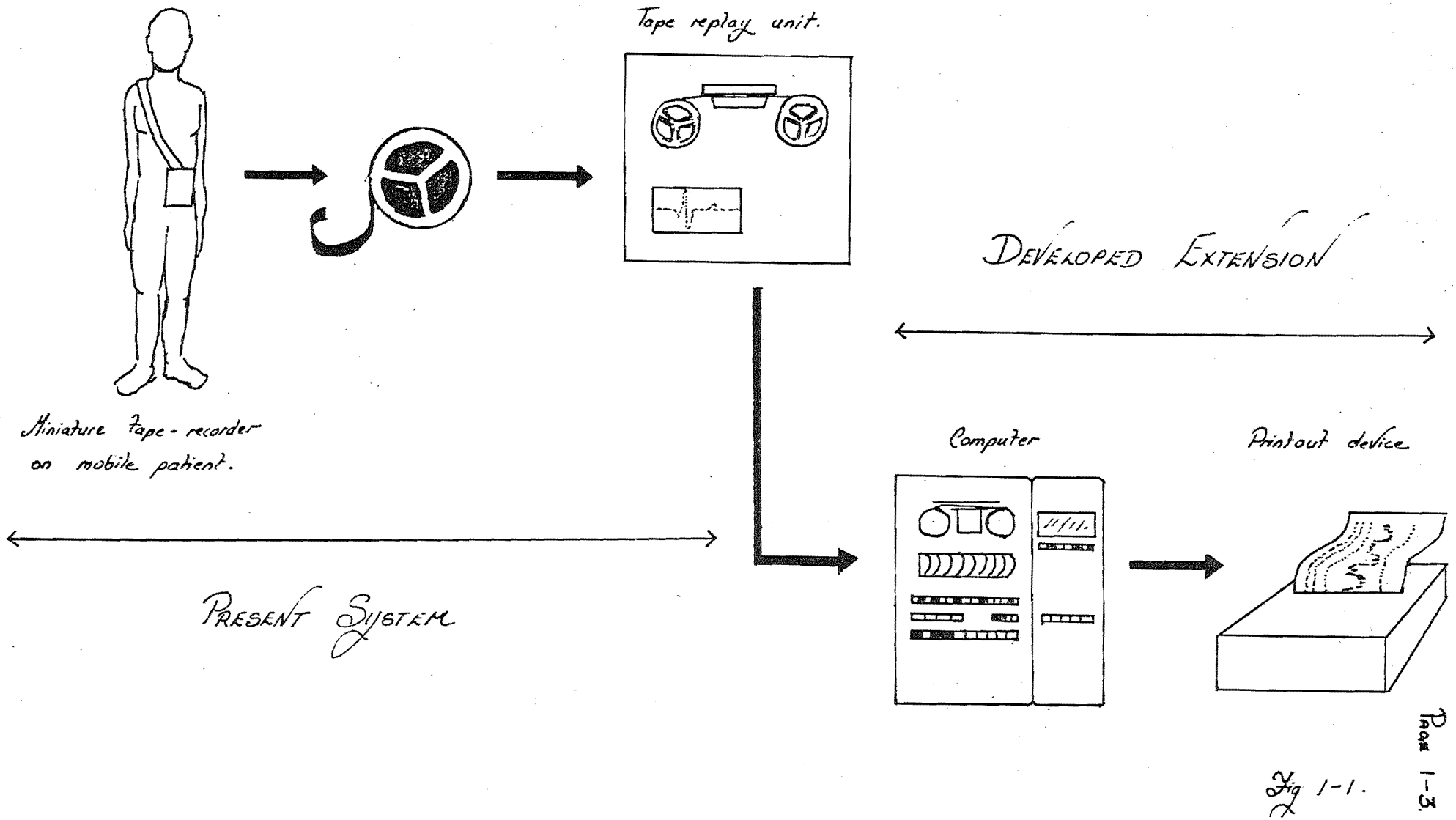

Tape replay unit.
Miniature tape-recorder on mobile patient.
DEVELOPED EXTENSION
Computer
Printout device
PRESENT SYSTEM

Fig 1-1.

also presented.

Chapter 2 reviews similar efforts that have been previously undertaken and chapter 3 details the development and theory of operation of the algorithm. The performance of the trial implementation of the algorithm is given and discussed in chapter 4. Chapter 5 proposes a full speed (ie 60 times real speed) implementation and chapter 6 concludes the thesis with a summary and discussion of the developed algorithm.

CHAPTER 2

Review of Current Analysis Methods.

Before starting to develop an algorithm to aid in the analysis of ECG waveforms, it is essential to be aware of any similar attempts that have been made so as not to reinvent the wheel if at all possible. Therefore a brief outline of the currently used manual analsis method wil be given, followed by a description of an analog method developed by J. M. Neilson in England. An outline of the AZTEC and ARGUS systems developed at Washington University is presented as is a general discussion of attempts to use spectral analysis. A system, developed at Stanford University, that uses correlation to perform the analysis is described and a final brief mention of other analysis techniques is also made.

2.1 Current Manual Analysis Method

2.1.1 Description of Method

The unit at present in use can play back the recorded ECG waveform in either real time or at 60 times the recorded rate. At the

higher replay speed hardware in the replay unit detects each beat and triggers the sweep on the display so as to overlay each beat on the previous ones. Thus abnormally shaped beats are shown as a background flash behind the overlaid beats.

The two replay speeds each have their own pickup heads with the real time head after the high speed head. This gives the operator time to stop the high speed playback after some important event has been noticed and to observe the same event in real time with the minimum amount of searching. The real time display to the operator shows the continuous ECG signal on the CRT so that the timing relationships between beats can be estimated.

If the operator wishes to permanently record an event, a strip chart recording can be made in the real time playback mode.

2.1.2 Advantages of the Manual Analysis.

A human operator has the ability to learn by experience and to apply the knowledge built up over a long period of time to the analysis of a particular tape. Also a person knows which events are of importance for a particular analysis and can mentally filter out those of no immediate interest. If any previous tapes of the same patient, or other patient with the same problem, have been processed then this information can also be mentally (and sometimes subconsciously) used.

The human brain is good at pattern recognition, even in the presence of high noise levels and is capable of rapidly assessing and classifying each beat.

2.1.4 Limitations of the Manual Analysis Method.

Manual analysis by this method requires a high degree of concentration by the operator as between 30 and 100 beats can be presented each second and retained by the persistence of the screen for about half a second. Concentration to this level is difficult to maintain for more than about five minutes at a time.

Because of the length of time required for an analysis, which can range from about 45 minutes for a good tape to over three hours for a tape with a lot of noise, and the relatively short concentration span of most humans, it is inevitable that significant events are missed.

2.2 Neilson's Automatic Analog Analysis Method.

2.2.1 Intended Function

Since the late 1960's J. M. Neilson of Edinburgh University has been developing an automated analysis system using analog techniques [1,2]. His method is designed to identify and count abnormal beats but no attempt is made to classify them.

2.2.2 Analysis method and Equipment Used (fig 2-1).

The method is based on extracting the integral of the difference between an incoming waveform and a stored template. The template is stored in a series of 'Sample and Hold' networks that can be sequentially 'read' by a comparison circuit.

The sequential reading of the template is initiated by a hardware pre-trigger and proceeds at a rate such that equivalent portions of the waveform are compared.

The incoming waveform and the sequence of stored samples are combined in a differential input integrator, thus providing the function:

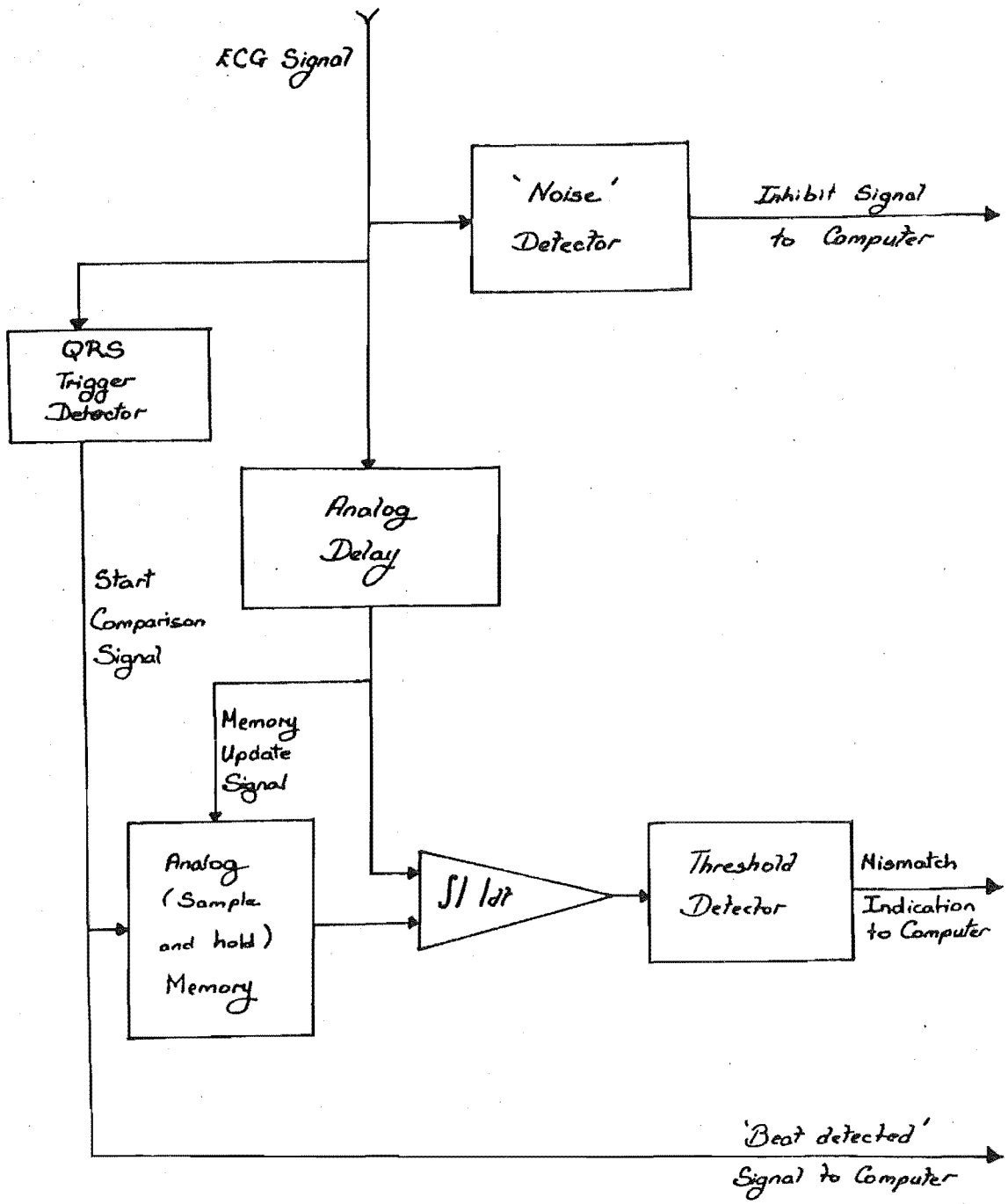
$$\int |f(t) - g(t)| \, dt$$

where f(t)=incoming beat

g(t)=stored beat.

If the waveforms are similar the result will be a low value and so a threshold circuit is used to decide between normal and abnormal waveshapes. When a normal beat is detected it is combined with the stored template to make up for losses in the sample and hold networks and to allow for gradual changes in the 'normal' waveshape.

Excessive noise is prevented from entering the system by a noise detector circuit which suspends processing for the duration of the



Neilson's Analog System.

Fig 2-1.

noise.

2.2.3 Discussion of Method.

While this method is useful if only the number of abnormalities is required, no indication is given as to the characteristics of any abnormalities encountered.

Being analog the waveforms are, theoretically, compared continuously but there is still effective sampling in the stored template and large numbers of sample and hold networks are cumbersome and expensive. This method also relies on the trigger supplying an accurate timing pulse. A small shift in time can result in a wrongly classified beat.

From available literature regarding this method [1,2], there is apparently no attempt to normalise the amplitude of the waveforms. This means that a beat can be wrongly classified purely because of amplitude variations.

2.3 A Digital Analysis Method (AZTEC/ARGUS)

2.3.1 Intended Function

AZTEC (Amplitude Zone Time Epoch Coding) is a preprocessing

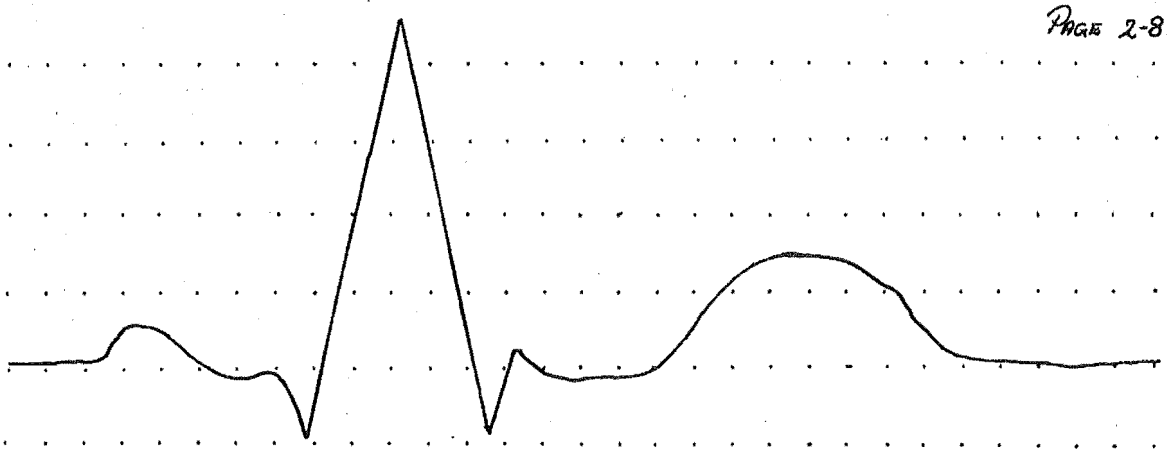
algorithm developed by workers at the Washington University's School of Medicine in the late 1960's to encode and compress the major features of an ECG waveform [3-5,11,31].

ARGUS (acronym unknown) and its successors, ARGUS/H and ARGUS/2H, are algorithms designed to use the output from AZTEC and produce a beat by beat classification of long term ECG recordings according to their clinical type and significance [3,4,11]. The major role of ARGUS is the real time detection of Ventricular Premature Beats in coronary care unit patients but ARGUS/2H is intended more for accelerated time, dual channel analyses. (The /H stands for high speed and /2H for dual channel high speed.)

2.3.2 Analysis Method and Equipment Used.

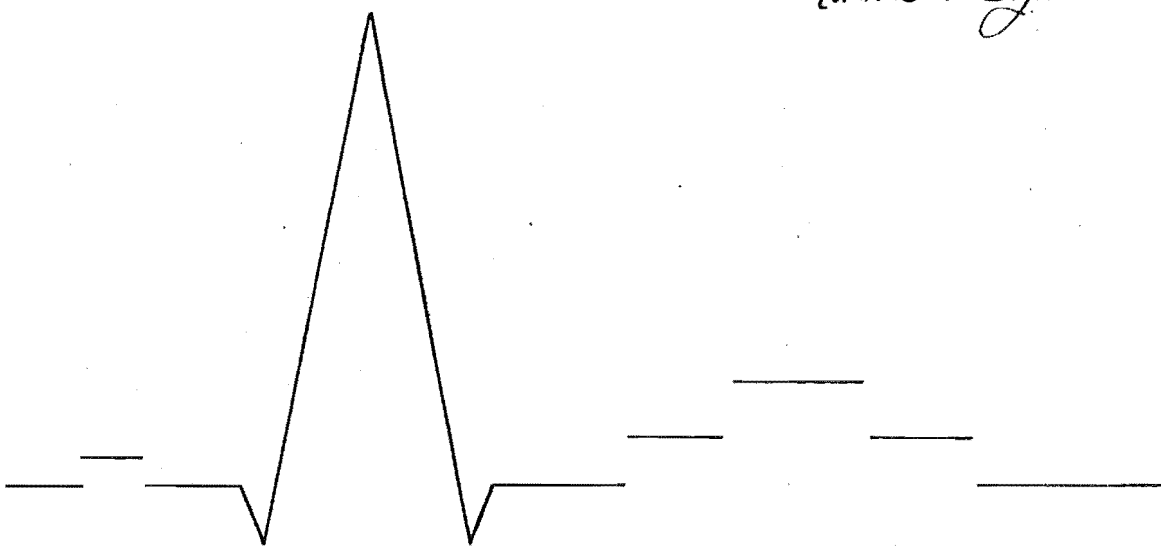
The AZTEC preprocessor takes in the digitised waveform and converts this into a series of straight lines (fig 2-2). The first step is to quantise the waveform and to note the time that the waveform stays within a particular quantum level. If the duration is greater than 8 mSec then the section is considered to be a line of zero slope and the duration and level of the line is stored. Lines are clipped to a maximum duration of 126 mSec.

If the duration of a quantised step is less than 8 mSec then the segment is considered to be part of a slope and the rate of change of the slope is recorded.



Analog Signal

Quantised Signal



Final Representation as Lines and Slopes

'AZTEC' SIMPLIFICATION OF ECG BEAT.

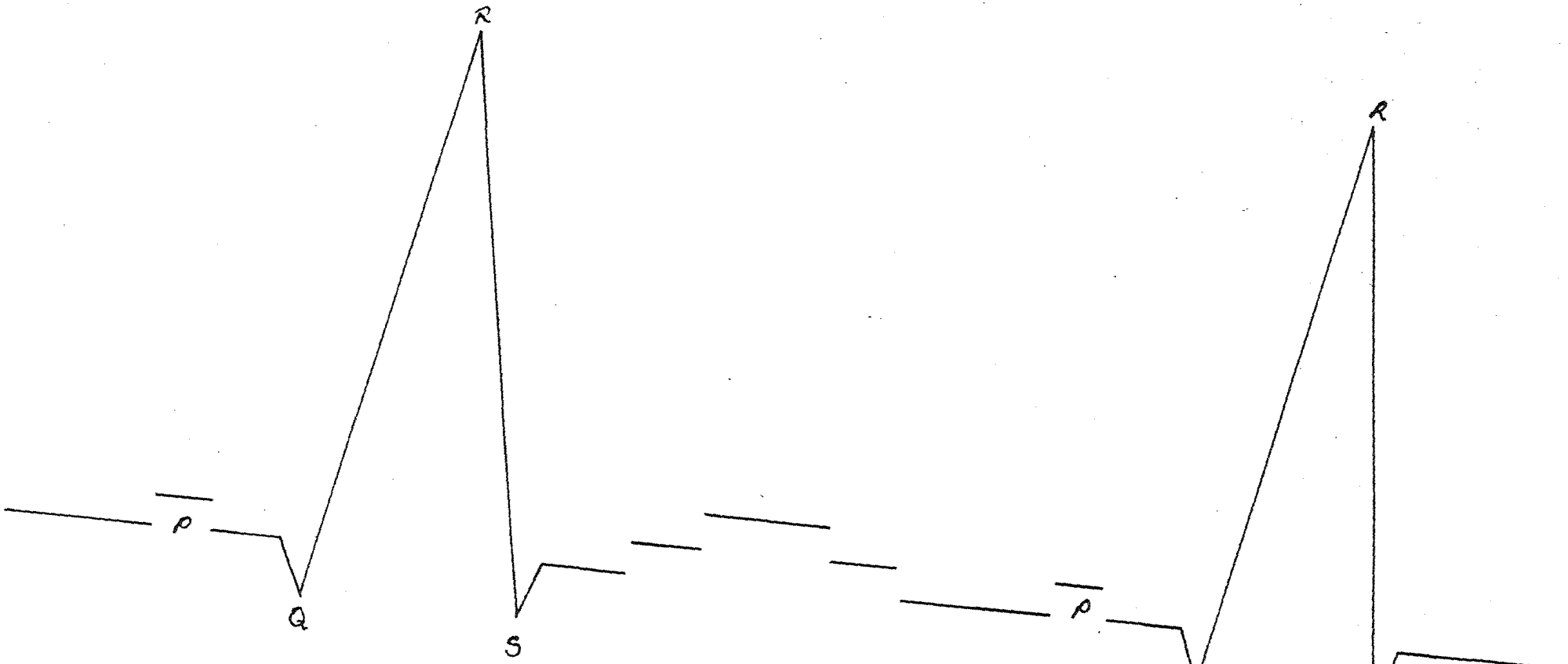
Fig 2-2

The second stage takes adjacent line or slope segments and combines them, if possible, into long, continuous lines and slopes of one of three gradients (small, intermediate and large) plus the direction of the slope. A 'Wiggle Subprocessor' then removes small slopes which have durations less than 100 mSec in otherwise flat regions. Thus the resulting waveform is a smooth, 'noise' free, characature of the original in which the QRS complex can be recognised by flat regions followed by large slopes of one sign followed by large slopes of the opposite sign (fig. 2-3). The deflections are classified as the 'R' wave and the 'Q' and 'S' points are then searched for around the R wave. If possible, a similar algorithm, working in parallel, is used to find any 'P' wave.

AZTEC then outputs the duration of the QRS complex, the Q-Q interval (ie the instantaneous heart rate) and, if a P wave has been found, the P-P interval, the duration of the P-R segment, the duration of the P wave and the Q-P interval. These are all standard measures of heart function that are used by cardiologists.

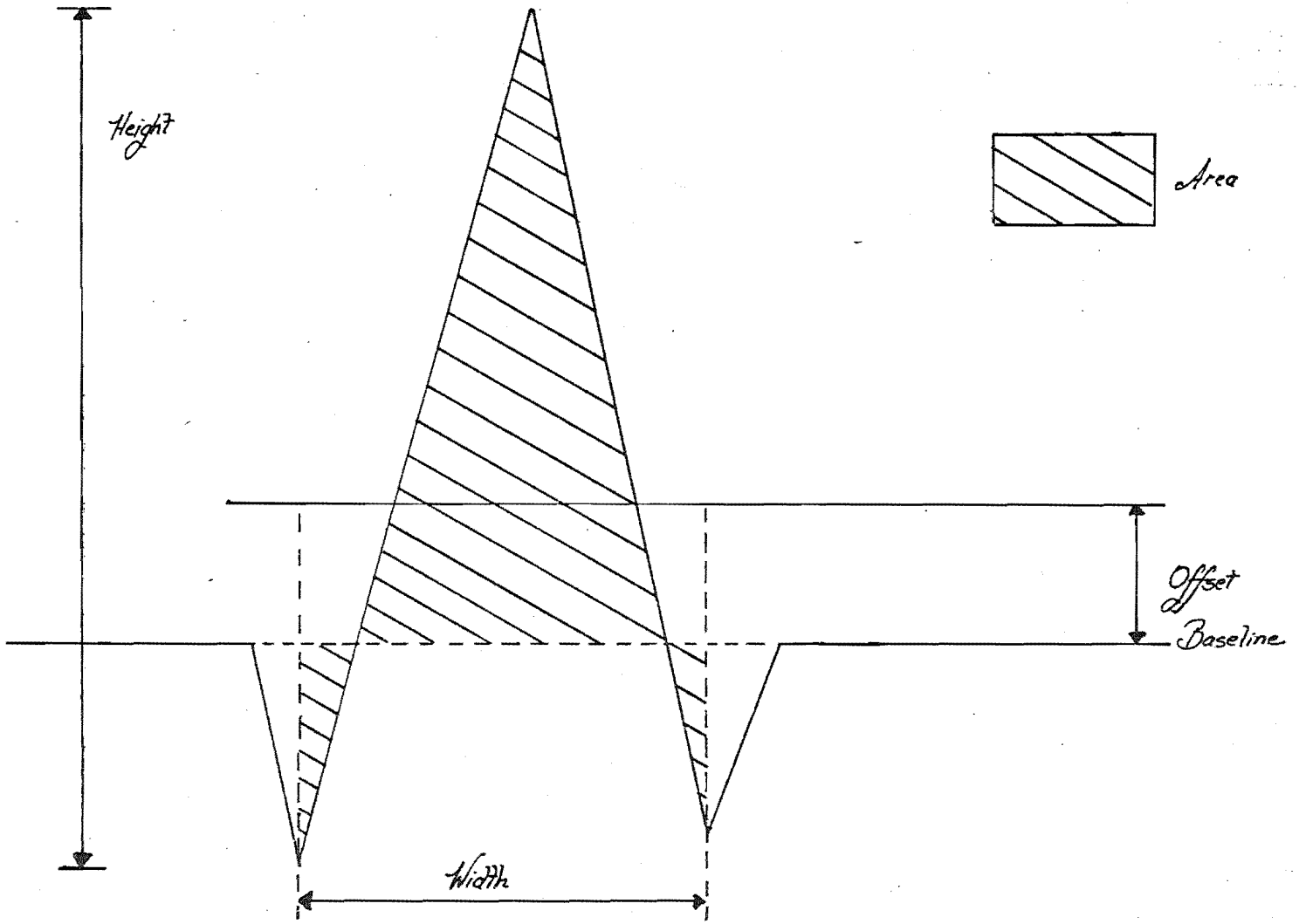
The preprocessor also supplies the height of the QRS and a baseline level to the following stage.

The QRS height and width are used directly by the ARGUS algorithm. In addition the QRS is quantified by the offset of the average height of the QRS from the baseline and the area between the QRS deflection and the baseline (fig 2-4).



FACTORS FOUND BY AZTEC

Fig 2-3



FACTORS FOUND BY ARGUS

Fig 2-4.

A family membership of a QRS complex is then determined based on clustering techniques in the four dimensional space defined by the above four characteristics of QRS height, width, area and offset. The beat is finally classified by considering the family it belongs to and the timing intervals to the surrounding beats.

ARGUS will output its classification of the beat along with a reconstruction of the ECG waveform from the AZTEC output. Any borderline beats are classified as such and, if processing is being done offline, are referred to the operator for final classification.

Result summaries of the overall heart rate and rates of occurance of the abnormalities are available at the end of an analysis period. More detailed results about each beat or run of beats can be requested at a later stage if required.

ARGUS/H has been implemented on an IBM System/7 to monitor two coronary care patients in real time, and ARGUS/2H has been implemented on two PDP11/34's working in parallel to analyse 2 channel 24 hour ECG tapes.

2.3.3 Discussion of Method.

ARGUS will attempt to classify each beat it comes to and looks in particular for Ventricular Premature Beats. However it does so by creating a large number of families ([31] shows at least 15) and appears to assign similar beats to different families. This means

increased processing due to the large number of possible family memberships.

Both forms of the ARGUS algorithm require powerful mini-computers in order to run, mainly because the algorithm attempts to classify each beat according to its clinical significance and tries to group beats into narrow family limits. On the other hand these algorithms have been running succesfully for several years in both online and offline applications.

2.4 Spectral Analysis Methods

2.4.1 Intended Function

When used on their own, spectral analysis algorithms [6-10,12] are used to detect ventricular premature beats in real time and to separate VPB's from normal beats on a beat by beat basis. The use of spectra can also be applied to QRS detection, especially in the presence of high noise levels [9,10] or, more generally, to filter out high and low frequency artifact from the ECG waveform to permit better operation of another analysis method [8,12].

2.4.2 General Analysis Method and Equipment Used.

At the centre of most of the spectral analysis methods is the Fast Fourier Transform (FFT) although use is made of narrow bandwidth

filters for some applications. The most common method is to compare the magnitudes of selected frequencies or groups of frequencies and use the fact that VPB's have wider QRS complexes. Thus there is a shift to the low frequency end of the spectrum [9]. Both methods produce a match coefficient for each beat and the beat is classed on this value.

The QRS complex has the highest frequency content of the ECG waveform, with the P and T waves having lower components, and thus the various portions of the ECG wavefrom can be separated by their frequency content [9,12]. This also allows for artifact filtering to be done, as studies [9] have shown that harmonics above 50Hz and below 0.5Hz can be rejected as noise. If required, a filtered ECG waveform can be reconstucted from the remaining spectral data for use in other algorithms.

Some other automatic analysis algorithms rely on accurately knowing the position of the QRS complex, and a spectral analysis technique is sometimes used to detect a constant reference point on the QRS [10]. Here again the fact that the QRS complex alone has the high frequency components is used and that there is a phase alignment of the higher frequencies centered on the QRS complex. A good estimate of the position of the QRS can therefore be quickly obtained wherever the QRS is within the measurement window (see appendix B).

2.4.3 Discussion of Method.

Although the FFT is faster to compute than a discrete Fourier transform, it still requires either a significant amount of processor time or complex and (at present) expensive hardware. Alternatively there are methods which only require two or three frequencies and the hardware and/or software requirements for these can be significantly less demanding, although it is difficult to select those frequencies that properly perform the required analyses.

Spectral analysis methods are sensitive to amplitude variations in the incoming wave form due to the use of the FFT algorithm, however this is usually reduced by comparing the ratios of frequency ampiltudes rather than the absolute data. The phase information given in the spectrum is not affected in this way and positional information about the QRS complex can be very easily obtained.

The shape of the ECG waveform is greatly affected by the position of the measuring electrodes with respect to the heart, but it has been found that the frequency magnitudes of the QRS are relatively insensitive to this variation [9]. This makes the use of standard templates more reliable.

2.5 Correlation Analysis Methods.

2.5.1 Intended Function.

The major work being done in the use of correlation techniques in automatic beat analysis has been undertaken at Stanford University.

The technique involves beat by beat analysis of ECG tapes in accelerated time and the placing of each beat into one of up to 63 families. The beats and their analyses are available for inspection by a cardiologist or a technician at the end of the analysis. The analysis primarily looks for Ventricular Premature Beats (VPB's, see appendix A).

2.5.2 Analysis Method and Equipment Used.

The Stanford algorithm has been implemented on an HP2100 series computer [13-16]. The ECG waveform is continuously sampled at 100 samples per second of real time and stored on disk. Thus any portion of the recorded beats can be reviewed at any time.

Each family created has 2 templates associated with it. Both are centered on the QRS complex, the first being 2 seconds long is for display to the operator and the second is 200 mSec long for the correlation calculations.

The algorithm automatically rejects beats that it considers to be noisy, based on the amount of movement of the ECG waveform prior to the QRS complex. It is generally agreed that this period is relatively free of features, excluding any P wave that might be detectable. If there is a large number of slope changes in this region the beat is considered to be artifact.

An incoming beat is correlated with the stored templates in the

order of the most recently found families and is placed in the first family for which the correlation coefficient is above that family's threshold for acceptance. If no such family is found the operator is requested to:

1. Force the beat into an exsiting family, lowering the threshold if appropriate

2. Create a new family using the beat as the template

3. Reject the beat as artifact.

A maximum of 63 families can be created, after which the operator is required to merge or delete those that are no longer required before the analysis continues.

After the shape classification has been determined, the timing of the beat with respect to the average of the several most recent normal timings is calculated and its degree of earliness or lateness recorded.

After the analysis, a series of reports is generated for the cardiologist. The first gives the statistics of the whole run and also for each template. The total number of beats detected, the number rejected as noise and the number put into each group is supplied along with the shape and correlation threshold of each template.

Secondly, R-R interval and family membership plots are generated with one hour of ECG per page. A similar but more condensed plot of the minimum and maximum heart rate for each 5 minutes of the analysis is also provided.

The third group of plots show the VPB rate against time and against the heart rate. This information is grouped into 15 minute segments. The first plot is used to see the relationship between the patient's activity and the abnormal heart beats, while the second shows any possible relationship between an increase in VPB rate with higher heart rates.

The frequency histograms for each family of membership against time are printed next, followed by ten randomly selected one minute segments of the stored ECG waveform giving group and timing values associated with each beat. This report is an attempt to demonstrate any systematic errors caused by bad template selection or correlation thresholds.

After viewing the above reports, a technician or cardiologist can request any portion of the recorded beats to be shown for further manual examination.

2.5.3 Discussion of Method.

One of the big problems with any automated ECG analysis scheme is that the size and shape of the ECG signal varies from one patient to

another and also in different recordings from the same patient. The latter situation is caused by different electrode sites on the body giving different waveforms.

The ability to digitise and save a complete 24 hour tape on disk allows analysis to progress at a rate determined by the algorithm and not the playback speed of the tape. It also allows for the post analysis display of any portion of the recorded beats for visual examination by a cardiologist. For these benefits approximatly 8.6 Megabytes of disk and magnetic tape storage are required.

Having up to 63 templates means that a large number of families with only slightly different characteristics can be created, each with a small membership. While the ability to merge or delete families is provided, each family to be merged has to be compared with up to 62 other families to determine the best match, thus making the human operator's decision tedious and demanding.

2.6 Other Methods.

Many methods for automatic ECG analyses, other than the general methods already described, have been proposed or tried, but few have found acceptance. Several attempts have been made to use statistical methods for the detection of particular types of abnormailties [17-19]. For example the Markov Chain principle [19] has been used to estimate what the next R-R period should be based on the past few

periods and an abnormality is detected when the expected period is not found. This method is especially suited to the detection of premature beats or missed beats.

Statistics have also been applied to the R-R interval data to detect transient (defined by the researchers as under 6 beats long) and persistent (over 6 beats) rhythms [18]. This method involves setting up a mathematical filter that corresponds to the required rhythm. When the output of such a filter becomes significant with respect to similar filters for other rhythms then that rhythm is recorded.

The last group of methods to be discussed introduces another technique of using various transfer functions on both the R-R interval data and the QRS shape. In Homorphic analysis [20,21] the transfer functions are selected in such a way that the desired features of the ECG waveform become prominent and thus easily detectable.

One group of workers has proposed the conversion of the features of the ECG waveform into a pre-defined alphabet [22]. Syntactic analysis methods developed for use in computer science then process the character strings to define and group each beat.

The above methods all tend to modify the information to a greater or lesser extent before proceeding with their analyses and therefore they involve more computation. They are also slow and so are mainly used of 'on line' or real time rather than accelerated time analyses,

and they all require large computers to run on.

The preprocessing, with the exception of some of the homorphic techniques, destroy the original information by data compression so that reconstruction of the original waveform is not possible.

CHAPTER 3

Development of the Algorithm.

Before developing an algorithm, a clear understanding of what the algorithm is required to do must be obtained. This initial step is described in section 3.1. The information available can then be examined to determine the possible methods of extracting the required data. The five approaches investigated in the development of this algorithm are described in section 3.2.

Section 3.3 gives details of the algorithm in its final form. A trial implementation of the algorithm made to test its performance is outlined in section 3.4.

3.1 Design Requirements

3.1.1 Sources of ECG Signal

The long term analysis system presently used at the Princess Margaret Hospital's Cardiology Department has been described previously (see section 2.1). Tape recordings of patients' ECG

activity are readily available. The tape replay unit can provide a continuous analog output of the recorded ECG signal and this analog output was taken to be the signal source available to the analysis system.

The time of occurance of a QRS complex within the ECG signal is supplied by the replay unit as a trigger signal coincident with the analog ECG signal. The replay unit uses this signal internally to overlay the successive beats (see section 2.1.1) during the accelerated time (60x) replay.

3.1.2 Information to be Displayed

It was decided early in the development of the algorithm not to attempt a clinical analysis of each beat as it was felt that a cardiologist or medical technician is in a better position to make a proper assessment of any abnormalities that might be present. To this effect an attempt has been made to summarise the information on a tape in order to highlight regions requiring complete analysis by a human operator.

The two important pieces of information needed for such a summary are the time interval between adjacent beats (called the R-R interval) and a shape classification of each beat. The R-R interval is a measure of the instantaneous heart rate and is measured from the previous QRS complex to the current one. This measure is important as many abnormalities have characteristic timing variations associated

with them (see table 3.1).

Shape classification is used to show when abnormalities have occured which cause the shape of the QRS to change, and to group similarly shaped beats together into families.

A combination of the R-R interval and the shape of a beat or sequence of beats enables a rough diagnosis of the patient's heart defects to be made, and typical regions of abnormalities found for further study.

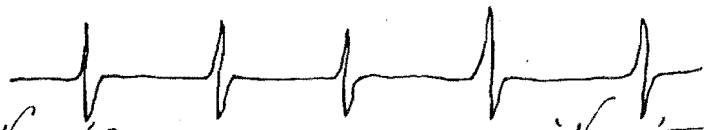
3.2 Waveform Analysis Methods Investigated.

3.2.1 R-R intervals Alone.

Table 3.1 shows some of the more common abnormal beats and rhythms that can occur. All of these abnormalities have characteristic rhythms and therefore the possibility of detecting an abnormality using R-R interval analysis alone was investigated.

A program was written that measured the time between QRS complexes, based on the hardware trigger from the tape replay unit. The time between the current beat and the previous beat was expressed as a percentage of the average period. This meant that any slow change in the heart rate would not grossly affect the calculated value but a sudden change would be shown by the percentage altering significantly from the 'normal' (100%) value.

SINUS

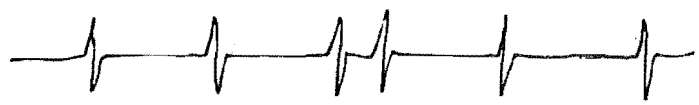


'Normal' Shape 'Normal' Timing

VENTRICULAR PREMATURE BEAT (VPB)

'Abnormal' Shape Short Interval followed by Long Interval

ATRIAL PREMATURE BEAT (APB)



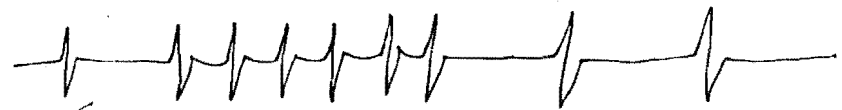
'Normal' Shape Short Interval followed by Normal Interval

PAROXYSMAL VENTRICULAR TACHYCARDIA (PVT)

'Abnormal' Shape Short Intervals Sudden Onset

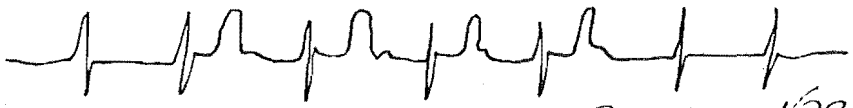
More than 3 'Abnormal' Beats in the Run.

SUPRAVENTRICULAR TACHYCARDIA (SVT)



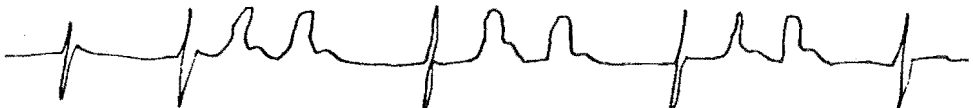
'Normal' Shape Short Intervals Sudden Onset

BIGEMINY



2 Coupled Beats Subset of VPBs.

TRIGEMINY



3 Coupled Beats Subset of VPBs

TABLE 3-1.

As it was not apparent how many beats should be included in the calculation of the recent average 'normal' R-R interval, four averages were calculated, 2,4,8 and 16 beats in length, and all R-R intervals were included in all calculations. Appendix F presents sample results from the program written to examine this method, and shows that the R-R interval alone is insufficient to provide adequate rhythm discrimination.

Because all measured periods, whether considered normal or not, were included in the average calculations, any abnormal R-R interval tended to bias the system's sensitivity to the opposite type of abnormal period. For example, an atrial premature beat (APB) will lower the average value so that, especially with the 2 and 4 beat averages, a following normal period can be detected as abnormally long. The 16 beat average was least susceptible to this, but it was unable to follow some of the exercise heart rate changes encountered.

The early/late percentage thresholds could be adjusted to be as sensitive as required, but it was found that beats with abnormal timing can occur within 5% of the normal R-R period and this is within the 10% variation observed for a properly functioning heart [23]. Also abnormalities sometimes show QRS shape changes with only slight R-R interval changes and these can give rise to ambiguous R-R interval sequences. Such beats are thought to originate elsewhere than at the Sinoatrial Node (see appendix E) and therefore the conduction path through the heart is incorrect.

On studying the results from the program it is difficult to find a satisfactory compromise between the use of long and short averaging periods. Long averaging periods (i.e. 16 beats) are best when looking for individual beats that are out of time but the average does not respond quickly enough to normal rate change, such as those due to excersise, thus producing false positives. Short averaging periods (i.e. 2 and 4 beats long) do not suffer this last problem but tend to track short runs of abnormal beats and so miss abnormal beats altogether.

The results obtained from the test program also showed the that this R-R interval algorithm can be defeated by sinus arrhythmia (see appendix A) as the variation in R-R intervals due to this normal characteristic can exceed +/- 20%, thus masking some abnormally premature beats [23].

In summary the analysis of heart beats by R-R interval alone can provide detection of gross timing abnormalities but some supplementary form of analysis is required to provide a more accurate classification of each beat.

3.2.2 Direct Feature Extraction

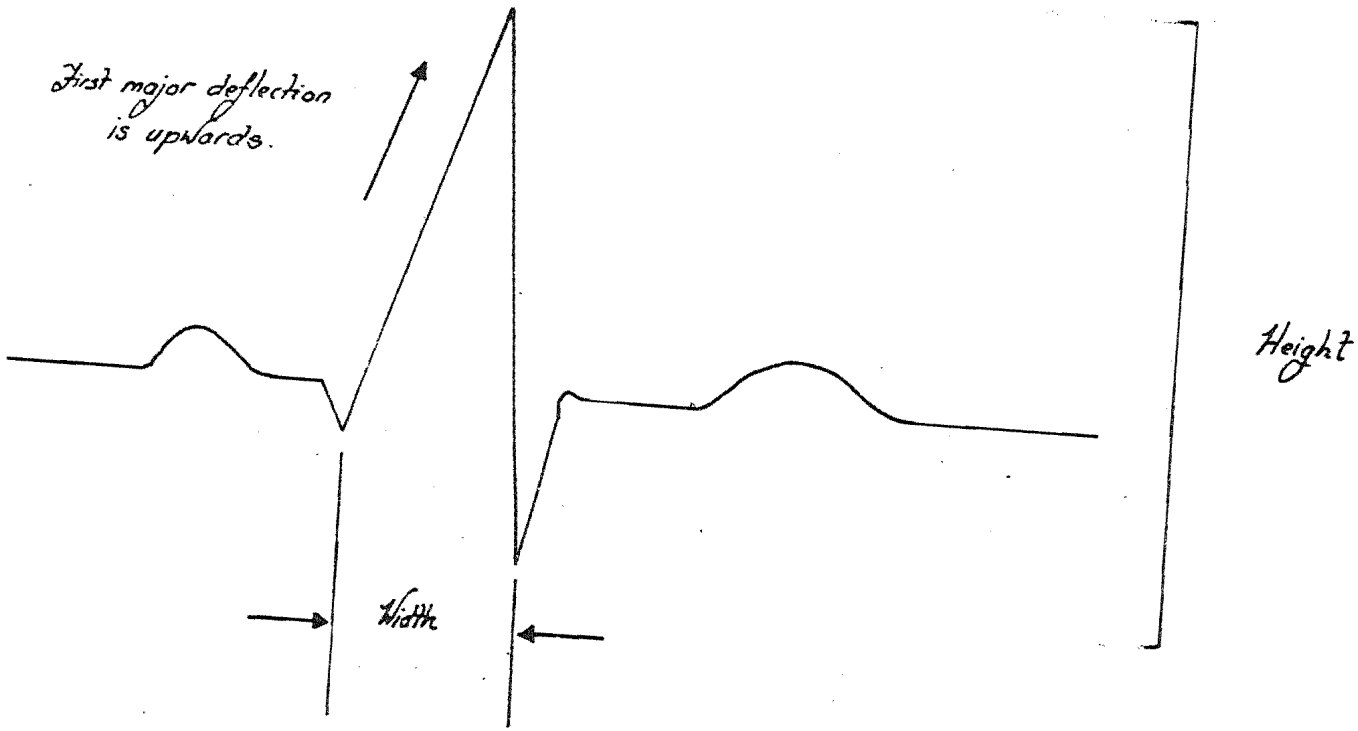
The results explained in section 3.2.1 demonstrated that the R-R interval alone was not reliable in detecting abnormal beats and rhythms and thus methods of examining the shape of the ECG waveform and extracting salient features were investigated.

Discussions with the cardiologists at Princess Margaret Hospital showed that the main features of interest in the QRS complex are the width and height of the QRS and the direction of the first major deflection (ie up or down) (fig 3-1). These features necessitated the finding of the onset and termination times of the QRS and the amplitudes of all maxima and minima between them.

The following algorithm was developed to locate the onset of the QRS by using the fact that the period between the P wave and the QRS is relatively smooth or flat compared to the QRS complex itself (fig 3-2).

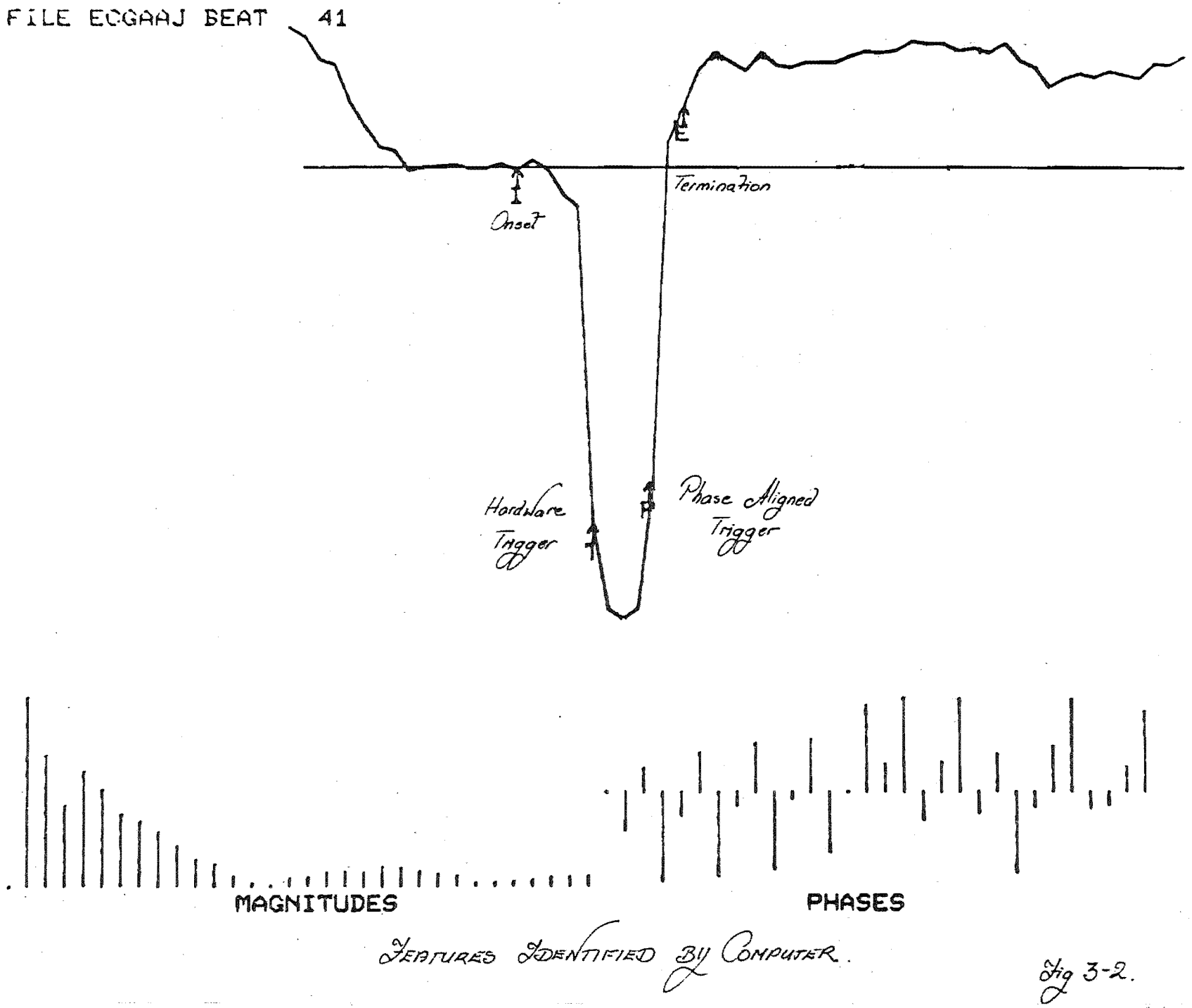
The hardware trigger point (T in fig 3-2) lies within the QRS complex and is usually within the first major deflection. Using the trigger as the starting point for the onset search, the values of the starting point and the three samples before it are averaged, and the sum of the squares of the differences between each of the four samples and the average is calculated. The first of the four samples used is moved one sample towards the start of the window and the calculation is repeated until a value is calculated that is greater than the pevious one.

The minimum value is retained and the above calculation is repeated but with steps in the opposite direction, i.e. towards the QRS complex, until the calculated value exceeds the minimum by a preset percentage. This point (I in fig 3-2) is taken to be the onset of the QRS complex.



MINIMUM SET OF FEATURES TO LOOK FOR.

Fig 3-1.



Features Identified By Computer.

Fig 3-2.

The performance of the algorithm was tested by comparing its results against the onset estimated by a human operator. In approximatly 95% of the estimates, the difference in onset point was +/- 1 sample and was always within + 2 to - 1 sample. The sampling rate used was 100 samples per second and thus this represents a variation in the onset position of approximately +/- 20 mSec. When compared to the width of the average QRS complex of 80 mSec, this variation in the onset position is significant.

To determine whether the algorithm was consistent within itself or whether the QRS complex was being under-sampled, a similar algorithm was tried on data samples at 1000 times per second. The variation between the computer selected onset and that point selected by the operator was found to be +/- 10 samples or similar to the slower sampling speed algorithm.

The P-Q segment of the ECG trace is generally accepted as being a time of no heart muscle activity and thus the onset of the QRS complex can be fairly well defined. However the same is not true for the S-T segment and an algorithm, similar to the one described above, to find the termination point exhibited wider variation than the onset algorithm (E in fig 3-2).

Due to the difficulty in finding the termination point and the inaccuracy in determining the onset position of the QRS, this feature extraction method was not further developed and no attempt was made to determine the direction of the first major deflection.

3.2.3 Use Of Spectral Analysis in Feature Extraction.

The failure of the QRS onset and termination algorithms described in section 3.2.2 to locate the limits of the QRS complex accurately, together with reports in several papers on spectral phase alignment techniques [10,24], led to an attempt to use spectral analysis methods to improve the location of specific points, such as the QRS onset, and to find a possible alternative to the hardware trigger in the QRS.

The hardware trigger indicated the approximate position of the QRS complex and a 64 sample window was set up so that the trigger was one third of the way in from the left hand edge. A 64 point FFT was calculated from the windowed data.

A phase alignment method of determining the position of the QRS, as described in appendix B, was implemented as a possible alternative to hardware triggering. The concept was to locate a point on the QRS that was in the middle of a major feature, thus accurately locating the QRS complex within the window and providing a restricted search region for the QRS onset. The phase alignment algorithm worked reliably (P in fig 3-2) as long as there was no other high frequency feature, such as another QRS complex or noise, within the window. However it failed when premature beats or high heart rates resulted in some part of the following QRS complex being in the sample window.

For this reason the use of spectral information was rejected, as it could not reliably be used as a software 'trigger' (and the

hardware trigger was found to be reliable) nor did it provide any extra information for use in the feature extraction algorithms previously discussed.

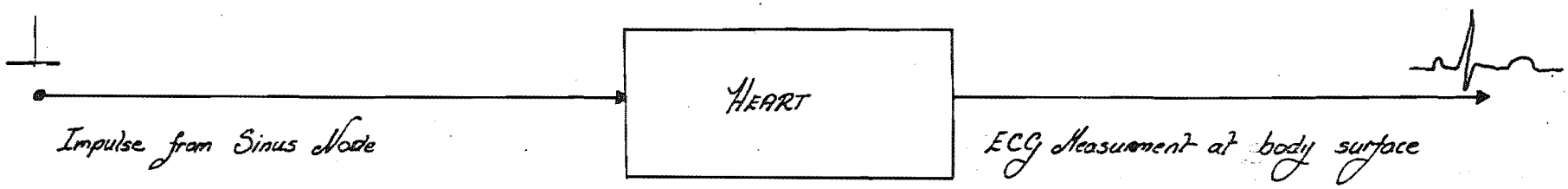
3.2.4 Digital Filtering

Since the feature extraction methods explained above have been shown to be unworkable, another way of determining the shape of the QRS complex was sought [12,25,26].

The shape of the ECG waveform can be considered to be the output of a filter (namely the heart) when excited by an impulse (fig 3-3). Thus it is possible to construct an inverse filter which takes the ECG waveform as its input and outputs a value corresponding to the 'goodness' of the match between the inverse transfer function and the incoming waveform.

With the increasing speed of digital hardware and the stability advantages of digital circuitry over analog devices, digital filters are becoming more practical for implementation in both hardware and software. The two types of digital filters are the 'recursive' and 'non-recursive'. Non-recursive filters combine the current and selected previous samples as weighted inputs to provide an output. Recursive filters also include selected previous output values as weighted inputs.

In this case the transfer functions can be found directly by



ECG AS THE IMPULSE RESPONSE OF A HEART 'FILTER'.

Fig 3-3

using a sampled ECG waveform. The next requirement was that the coefficients of the inverse filter must be easily obtained, and this indicated the use of non-recursive digital filters [26], as the input weighting coefficients are the sample values of the required transfer function's wave shape. More simply, the answer is to use a correlation process.

Thus the use of correlation functions were investigated as outlined in section 3.2.5.

3.2.5 Template Matching

As explained in section 3.2.4, use of the correlation technique was considered to be the best method of determining the shape of an ECG waveform and the QRS complex in particular. However the use of the correlation technique requires one or more waveshapes to be stored for comparison with each beat analysed.

Because of patient to patient variations in the shape of the ECG waveform, as well as the variations due to differing electrode placements, no standard ECG waveform can be used as the 'normal beat' template. The same argument applies to any abnormal beats that might be recorded from the patient. For these reasons the templates need to be drawn from the recorded beats themselves.

Typically patients have one predominating ECG wave shape which is used for the normal beat shape template. Most patients have a maximum

three different and abnormal beat shapes which must also be stored, so the number of templates required is not as large as may first be thought.

Various factors, such as slight movements of an electrode away from the skin surface, can vary the amplitude of the ECG signal which will cause the output value of the correlation function to also vary. Thus the correlation function must be normalised with respect to both the stored templates amplitude and the amplitude of the analysed beat. Another factor that can affect the resultant value of the correlation function is any 'baseline wander' or variable DC offset in the recorded signal as any DC offset acts as a scaling factor on the resultant correlation function value.

To eliminate these effects the correlation coefficient becomes:

$$\frac{\Sigma f(t).g(t)}{\sqrt{\Sigma f^2(t).\Sigma g^2(t)}}$$

where f(t)=F(t)-F(t)=the ECG waveform

and g(t)=G(t)-G(t)= the stored template.

If the incoming beat and stored templates were represented by 16 samples, then the dividend of the above formula requires 16 multiplications and 15 additions for each template. The divisor involves only 16 multiplications and 15 additions per beat as the template normalisation factor is a constant. This was considered to

be a significantly large number of operations and so methods of reducing the computational requirements were investigated.

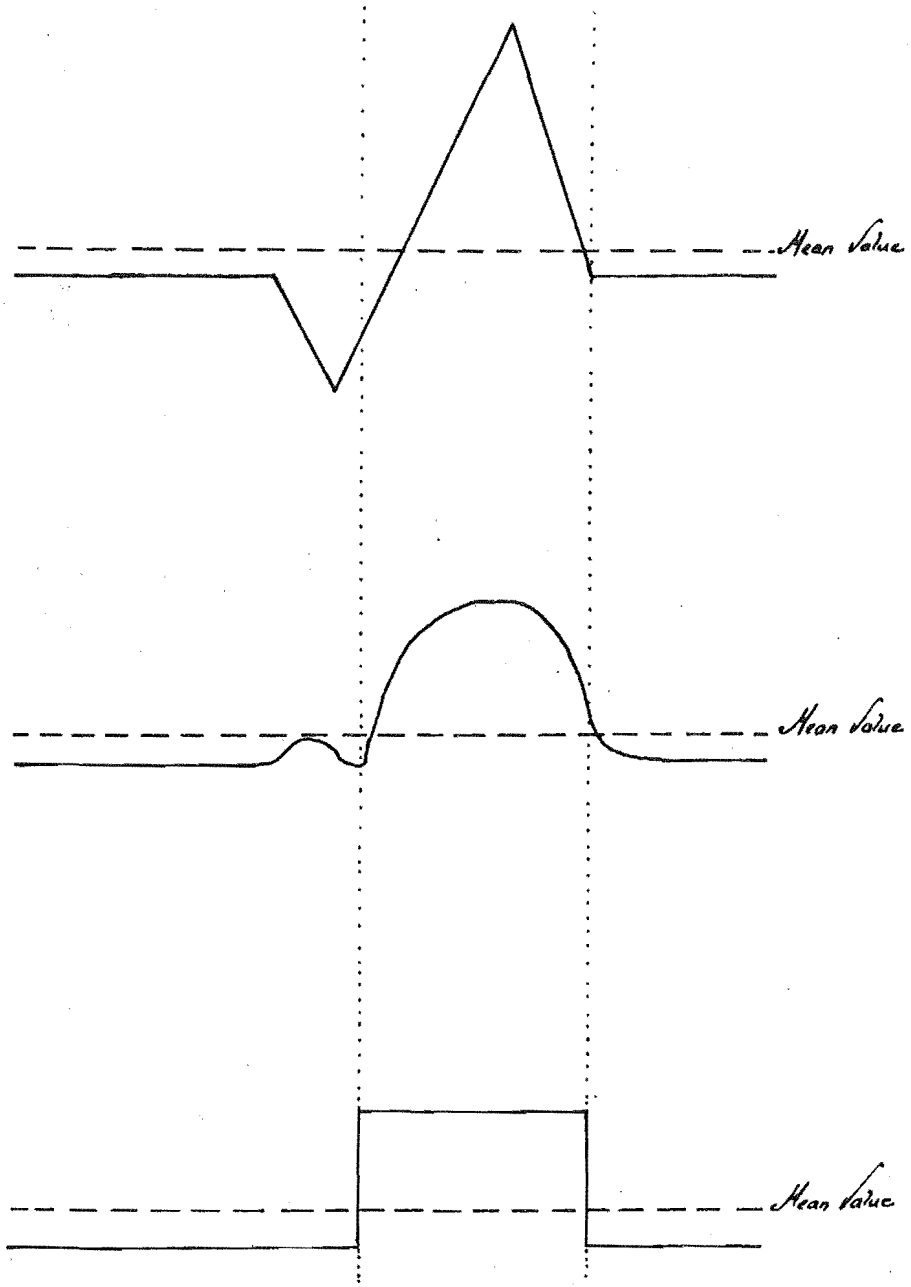
A paper on the use of Basis Functions [24] led to the idea of reducing the templates to a two state format by retaining only the signs of the original data values about their mean. Thus the correlation function is reduced to adding or subtracting the sample values of the incoming waveform according to the sign of the stored template values.

One advantage of this scheme was that the normalisation for the template always had the value of 16 (as the template was 16 samples wide). However it was realised that it was possible for two similar but clinically different waveshapes to have the same modified template shape, and thus the abnormalities could not be separated (fig 3-4). Too much amplitude information was being lost by the template approximation.

Further studies into the above short form correlation equations showed that it was as fast for a computer to multiply two numbers as it was for it to add them, and so the full correlation formula was favoured over any modified algorithm.

3.3 Algorithm Details

The analysis method finally implemented combines the R-R interval



Two Dissimilar Beats Producing the Same Modified Template.

Fig 3-4.

analysis method described in section 3.2.1, and the template matching technique, which was introduced in section 3.2.5., to detect abnormalities.

3.3.1 Interval Analysis.

The sample used to time the QRS is the first sample after the hardware trigger indicates it has found a slope of sufficient magnitude to qualify as the start of a QRS. As the waveform is sampled at a 100Hz rate, the R-R interval, in milliseconds, is ten times the number of samples since the last trigger mark.

Another measure is formed from the R-R interval that expresses the interval as a fraction of the current expected R-R interval using:

RR = actual R-R interval/expected R-R interval

The resulting fraction is compared with two thresholds, initially set to 0.8 and 1.2 but adjustable by the operator, and the beat is classed as early, on time or late depending on the results of the comparisons.

If the beat is classed as 'on time' its interval is included in the calculation for the next expected interval. The interval is calculated by the exponentially decaying form:

$$\overline{RR}_{n+1} = \overline{RR}_n \cdot (1-\alpha) + RR_n \cdot \alpha$$

where (α)=0.24 which is chosen to approximate a six beat average.

The exponentially decaying form used was chosen over a

conventional six beat average because the latter required either summing the previous six intervals each time or correcting for each sample when it was no longer to be included in the calculation, both of which required the storing of the six values used. The equation used above only requires the storing of the expected interval and the inclusion of the last interval found. Previous values automatically loose their influence in successive calculations without being explicitly removed.

When the beat is either early or late it is not included in the average as doing so was found to improperly bias any following 'on time' beats (see section 3.2.1).

The above algorithm may fail on encountering a sudden rate change or a very noisy section of recording.

The running average is able to track normal increases and decreases in the heart rate due to the requirements of the body but does not follow any sudden, abnormal, change such as occurs in tachycardia and bradycardia (see appendix A). Although tachycardia and bradycardia are abnormal and should be signaled as such, the normal rhythym following such changes may be significantly different from the normal rhythym preceeding the episode, thus possible giving rise to another abnormal run being incorrectly indicated.

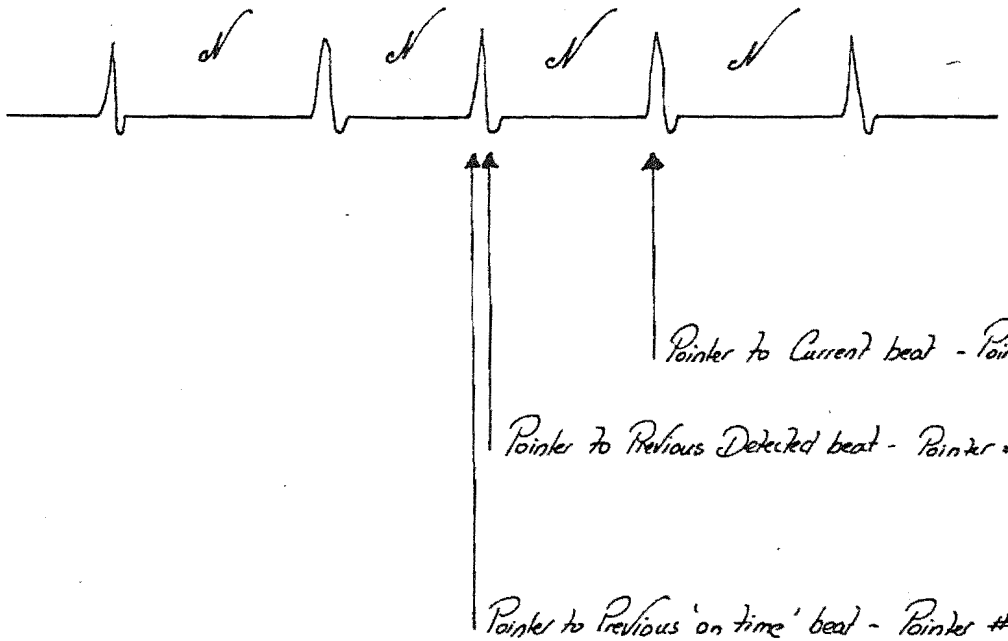
Recorded noise can last long enough for a normal heart rate change to be masked and a situation similar to the above can arise.

The solution to this problem is to detect when a rate change occurs, reset the current average heart rate to the new rate, and continue from there. This solution, however, allows an isolated premature or late beat, or a short burst of noise to cause resetting of the average when it is not desirable.

The final method developed required 3 pointers to properly perform the R-R interval analysis (fig 3-5). The first points to the trigger point in the current beat and the second to the trigger point of the previous beat. The difference between them gives the current R-R interval. The third pointer points to the last 'on time' beat detected. Usually this is also the previous beat, but under the conditions mentioned above it remains pointing to the last 'on time' beat.

When an 'on time' beat is detected, the second and third pointers are reset to the position of the first, ready for the next R-R interval determination.

If a noise burst or a sudden rate change occurs, the third pointer is left at the beat preceding the change (fig 3-6). At the end of each R-R interval determination the distance between the first and third pointers is compared to the average R-R interval. If the distance between the pointers is more that 3 times the average interval, a possible rhythm change is said to have occured and the average R-R interval is reset to the last R-R interval. The value of 3 times was chosen to allow for isolated ventricular premature beats



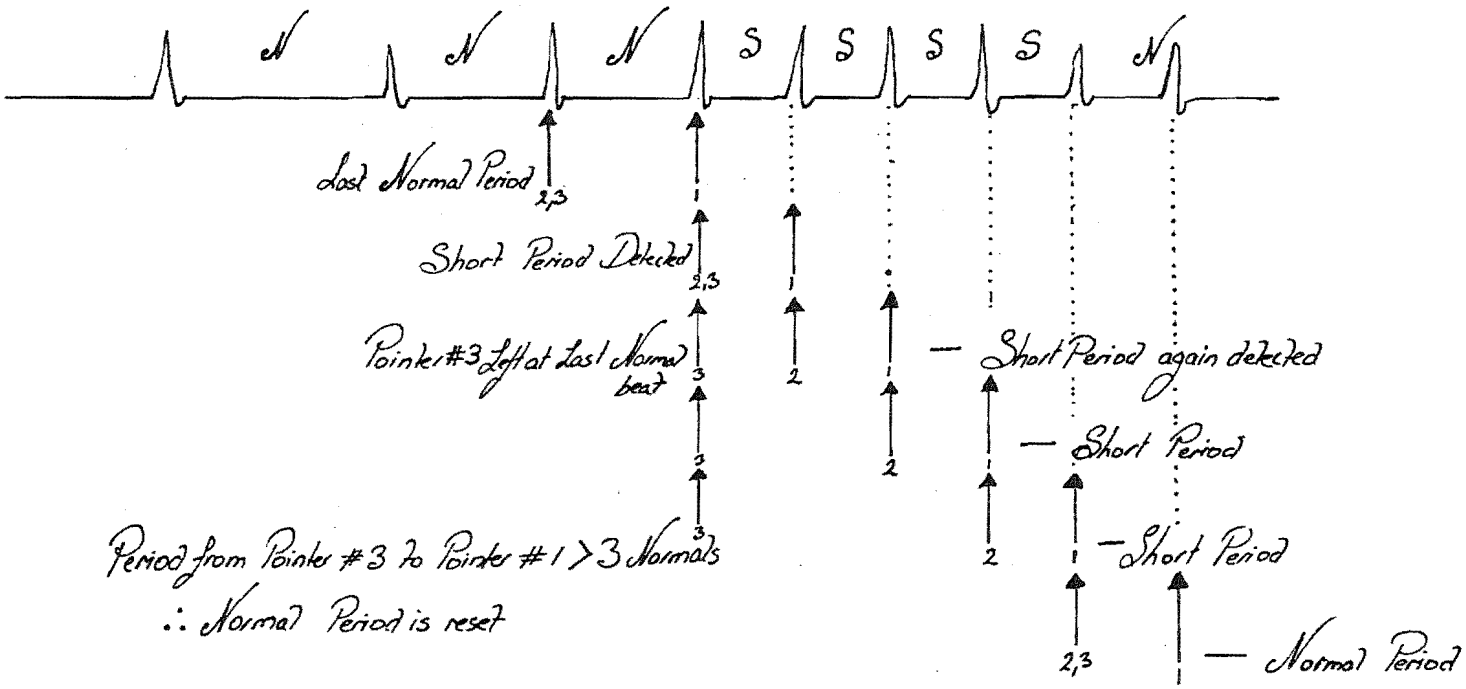
N - Normal (on time) R-R Interval

S - Short R-R Interval

L - Long R-R Interval

Pointer Locations for Sinus Rhythm

Fig 3-5



POINTER LOCATIONS DURING ENTRY TO TACHYCARDIA

Fig 3-6.

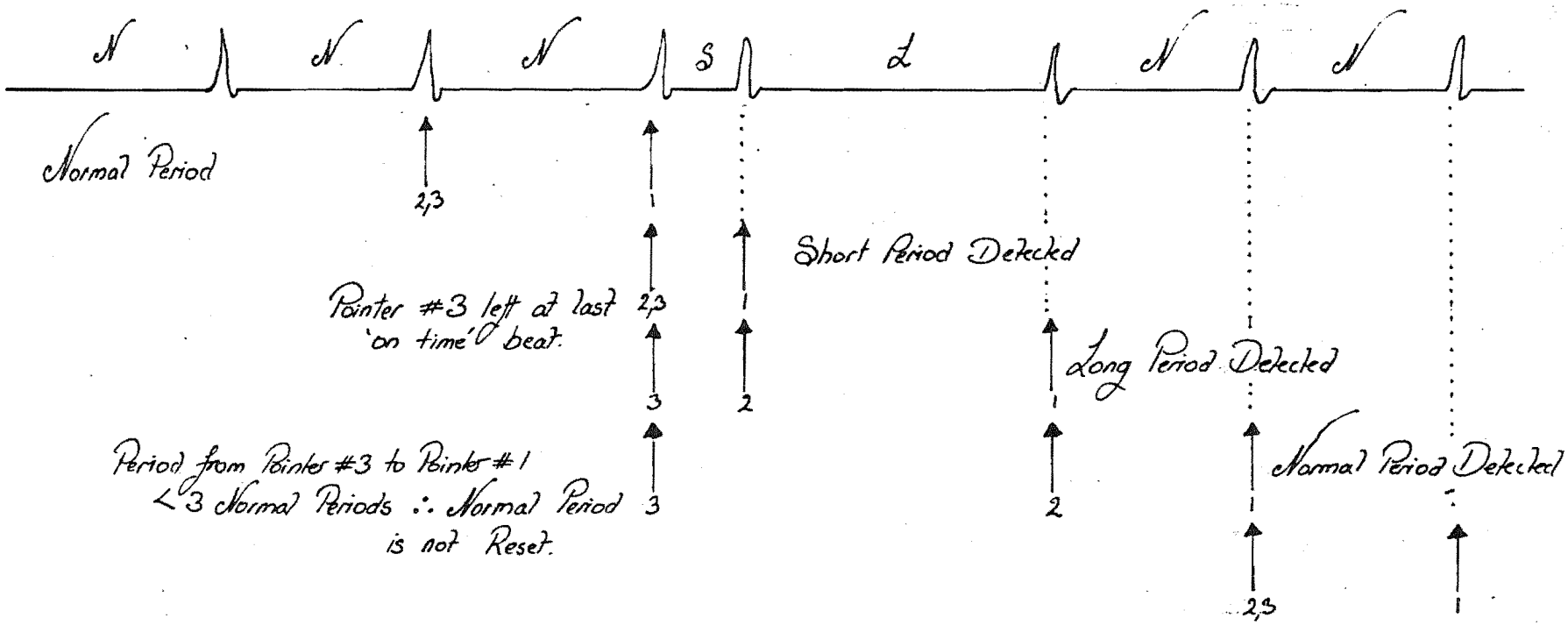
(VPB's) (whose total effect lasts for 2 R-R intervals (fig 3-7)), and yet to respond quickly to a run of beats at a different rhythm. If the resetting process occurs in the middle of a noise burst then the average will be reset every few trigger points until the noise is passed.

In order to start an analysis the first beat is used only to provide an initial timing reference. The second trigger is used to calculate the instantaneous R-R interval in the manner described above but the value is also used to preset the running average.

3.3.2 QRS Shape Analysis

The intention of the algorithm is to characterise the shape of an ECG beat by looking only at the QRS complex. Therefore a window, 20 samples or 200 mSec wide, is placed around the QRS complex so the trigger point, which defines where the QRS is, is the sixth sample from the left hand edge. The average QRS is 80 mSec wide but abnormally wide complexes can extend to 130 mSec and will thus still be inside the window.

The correlation process described in section 3.2.5 and used in the shape analysis of each beat, requires the average of all the 20 samples of the beat to be calculated and then subtracted from each sample in order not to correlate any DC offset present on the signal. The resulting values are used in the subsequent correlations.



Pointer Positions During VPB

Fig 3-7

The algorithm requires a set of templates against which the shape of a beats QRS complex is compared. These templates are the beats that have been declared by the operator to be representative of an abnormality that the patient has.

The templates are used in the following manner. A template is defined to be 16 samples long with a trigger point as its fourth sample. The trigger points of the template and the incoming beat are aligned so that equivalent portions of the QRS complex are compared.

The hardware trigger is sensitive to amplitude changes in the incoming analog waveform, and so the best match between the beat and a stored template may not be with exact alignment of the trigger points of the two wave shapes. Experience has shown the variation of the trigger position due to the hardware trigger to be within +/- 10 mSec which corresponds to +/- 1 sample of the digitised waveform. There is also a small error of up to 1 sample as the trigger position is quantised to the sample after the hardware trigger pulse was detected. To compensate for these errors, the dividend is also calculated with the trigger points of the beat and the templates offset by +/- 1 and +/- 2 samples and used to calculate a set of intermediate results. The largest of the intermediate results for each template so obtained is used to complete the calculation, thus substantially reducing the amount of computation required. This shifting is the reason for the storage of 20 samples per beat.

Recalling the correlation coefficient equation used (see section

3.2.5): $$\frac{\Sigma f(t).g(t)}{\sqrt{\Sigma g^2(t)}}$$

the dividend is calculated for each template (g) in use at that time. Because the various templates can have different amplitudes, the above partial results must be normalised by the:

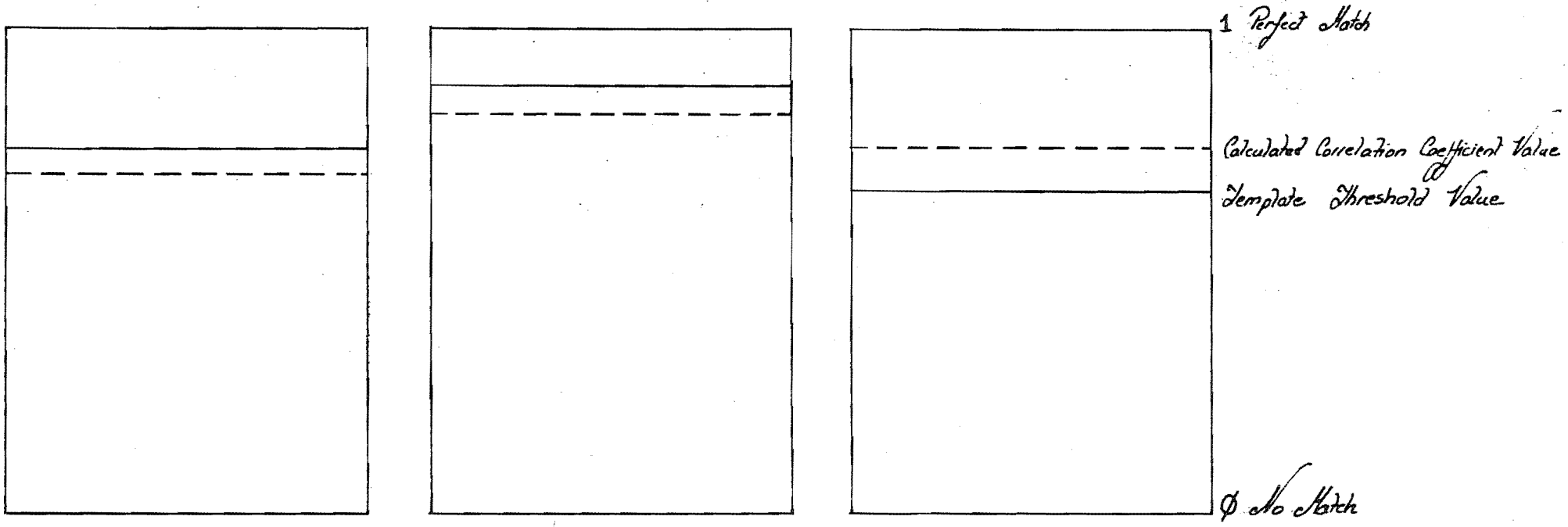
$$\frac{1}{\sqrt{\Sigma f^2(t)}}$$

term before comparisons can be made between templates. As this term is constant for a particular template, it is stored when the template is created and does not require recalculation.

Thus it is possible to order the templates from best to worst match with the beat. The next step is to see if the match is good enough. This requires the calculation of the beat normalisation term to remove amplitude variations between beats.

The probability that similar beats are identical is close to zero and so some variation must be allowed for. This is done by associating a threshold value to each group (fig 3-8). A beat's correlation coefficient with a template must be above that template's threshold before the beat can be included in that group.

Each template is tested in the order of best match and if the correlation coefficient is above a template's threshold then the group count for that template is incremented and the search for a match ended. If it is not above the threshold then the next template is



Beat included in group #3

TEMPLATE MATCH SELECTION

Fig 3-8.

tried until all the templates have been tried and none of the coefficients are above the corresponding thresholds. At this point the algorithm calls for the assistance of an operator, as outlined below, before carrying on to the next beat.

In summary each template has associated with it:

1. An auto correlation value for use in normalising with respect to the template.

2. A shift count which shows how much shifting of the beat with respect to the template was required to give the best match.

3. A correlation coefficient threshold that must be exceeded before a beat can be accepted into that group.

4. A beat count of how many beats have been included in the group.

5. A temporary correlation coefficient of the last beat with the template.

In the case of manual intervention the options available to the operator are to force the beat into a group, to start a new group or to reject the beat as noise. The first option is mainly used when a coefficient marginally fails to cross its group's threshold, and the operator considers that the beat and the template have the same features with regard to the information required from the analysis.

The threshold of the template is lowered to the value of the coefficient or to a preset bottom value, whichever is higher. Normally the preset bottom level is at such a value that another template should be used or created. See appendix D for a discussion of the effect of the threshold value.

If the operator decides that no existing template properly matches the beat, but that the beat is of importance, then another template can be created using the beat as its waveform, and a default threshold set for it. Thus the waveforms used to test the following beats are based on the patients own QRS shapes.

The third option open to the operator is to reject the beat as noise. This occurs when noise has produced a false trigger or has corrupted the shape of a QRS complex such that it could not be properly grouped by the operator. If this option is taken the beat is considered to not exist and no further processing is done on it. Also it does not enter into the timing analysis and so the next beat's R-R interval is from the last non-noise trigger.

Because of the use of a patient's own beats as comparison templates, before any analysis can begin the operator needs to scan the tape to select a normal beat that may be used to set up a normal template which is always given the sequence number #1.

3.3.3 Beat Classification

As it is not the intention of the algorithm to provide a clinical description of each beat, but rather to provide a summary of the events that were detected as significant in some way, the final decision making process is very simple.

The information supplied is an 'On time / Early / Late' indication from the timing section and a group membership number from the shape analysis section. A normal beat is one which is both on time and belongs to group #1. Any other beat is flagged as abnormal and an indication is given as to the reason(s) for being abnormal i.e. late and/or abnormally shaped.

This summary of the events on the tape recording then shows a technician or cardiologist where to look on the original tape for more information about a particular feature.

3.4 Algorithm Implementation

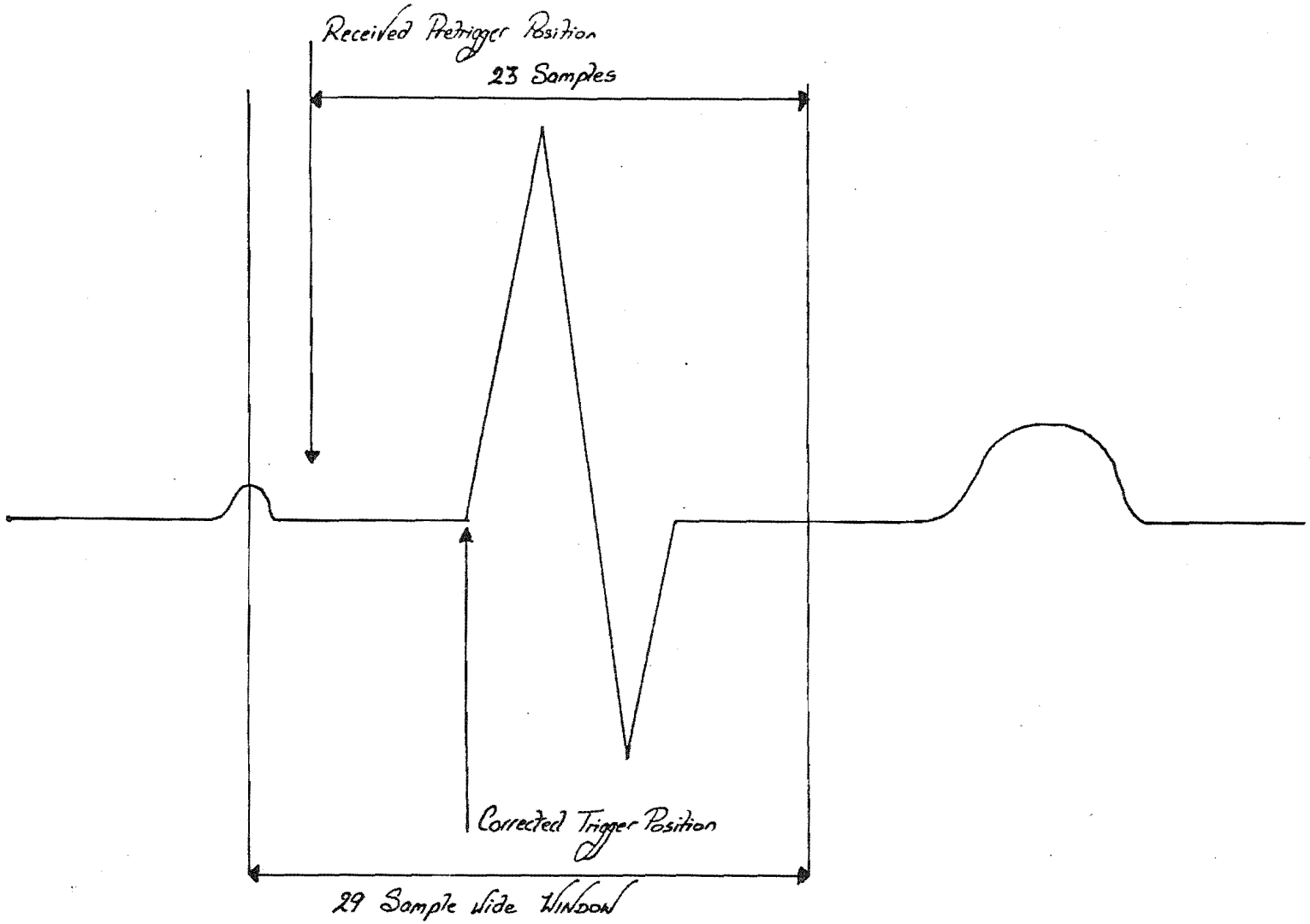
The algorithm described above was implemented on a laboratory minicomputer in order to test the principles and evaluate the summary of results of the algorithm. The next three sections respectively cover the inputing of the ECG signal from the tape replay unit to the computer, the programing on the computer of the algorithm itself, and a description of the display methods used to output the algorithm's results.

3.4.1 Input Conversion and Processing

The routine work of the Cardiology Unit provides a wide selection of 10 and 24 hour ECG tapes and these are used as an input signal to test the performance of the algorithm. Because this implementation of the algorithm is to be used mainly for testing and evaluating the algorithm's performance, no attempt has been made to make the algorithm operate at 60 times real speed. Therefore the signal is input into the computer, digitised, windowed around the QRS complex and stored on a disk file together with some timing information. The algorithm is then able to accept beats whenever required, with no time restraints.

The input sampling routine is written to operate at 60 times real speed as the ECG tape playback speed cannot be altered. The sampling routine receives a series of samples from the analog to digital converter (ADC) which is a peripheral on the computer used, and also a digital pre-trigger to indicate when a QRS complex is about to start. Every sample is placed into a ring buffer in memory so that samples of the waveform before the trigger can be used. The program counts 23 samples from the ADC after the pre-trigger is received to correct for the early arrival of the trigger signal and places the last 29 samples, together with the sample count from the start of the file of the corrected trigger position, into the disk file (fig 3-9).

When sufficient beats to allow proper functioning and testing of the algorithm have been received, the disk file is closed and made available to the analysis algorithm.



Sampled Window Around QRS Complex

Fig 3-9

3.4.2 Programing the Algorithm.

The algorithm described in section 3.3 is embedded in a control structure which allows the main variables and intermediate results to be examined and altered for experimentation.

When the algorithm is ready to accept a new beat, it recovers the next set of samples from the disk file, subtracts the mean value of the samples from each sample and passes these with the timing value to the analysis sections of the algorithm.

The output results from the algorithm, the R-R interval, the beat's group number and any early/late indication, are stored on another disk file and also are displayed on the operator's graphics terminal. The saving of the algorithm's output on the disk file enables the redrawing of selected portions of the output after the analysis is complete.

3.4.3 Description of Output.

The algorithm requires two output display formats; one for the analysis results and the other for providing the operator with the necessary information to direct the algorithm.

The analysis results display is divided into 3 main sections. The first of these (section A in fig 3-10) gives information about the

patient and the date and type of recording analysed. The second section (B in fig 3-10) contains the abnormal beat indicators. The top line is always present and shows early beats by a vertical bar below the line, and late beats by a vertical bar above the line. Immediately below this are a series of lines, one for each abnormal template created in the analysis process (there are two in fig 3-10 and three in fig 3-11). A vertical bar on one of these lines shows a beat has been matched to that template. A beat does not produce any such bar if it has been classified to be of normal shape.

The third section (section C in figs 3-10 and 3-11) is the R-R interval plot. The x-axis is time and can be varied to allow for different amounts of definition. The y-axis shows the time in mSec between each beat and the preceeding one.

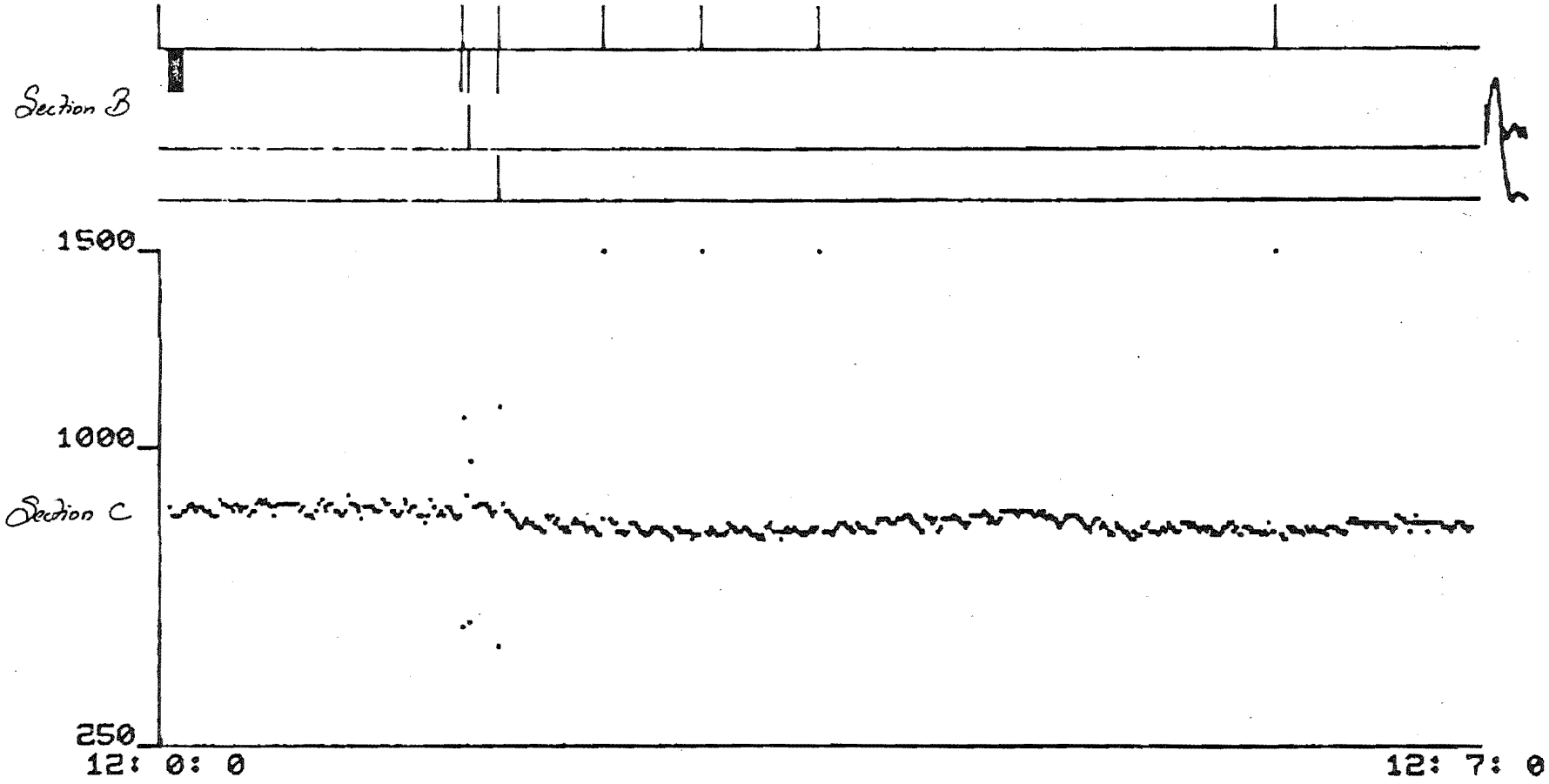
Figure 3-12 shows the output format used for interaction with the operator. This format is used whenever the algorithm fails to classify a beats shape and requests a decision from the operator.

Section A of fig 3-12 shows the up to six currently defined templates, the first being the 'normal' and any others being abnormal templates previously defined by the operator. Below each template drawing is the sequence number used to identify the template, and the threshold, correlation and shift values of that template. (The meanings of the threshold, correlation and shift values are given in section 3.3.2).

PRINCESS MARGARET HOSPITAL CARDIOLOGY UNIT

BEATS IN FILE: 512 STARTING TIME:12: 0: 0 END TIME:12: 6:56

STARTING DATE: 1-Mar-80



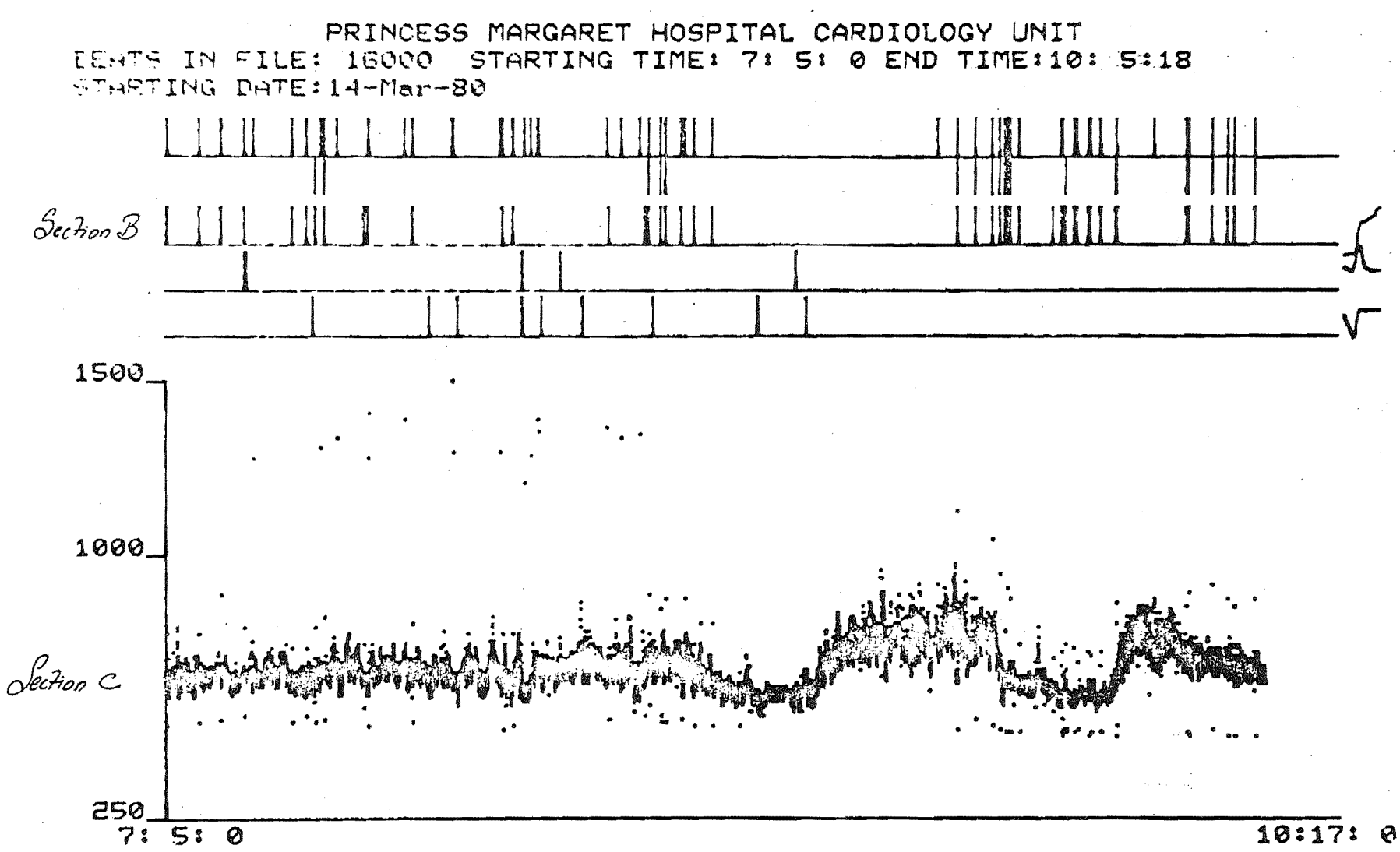
Section A

NAME:S E AGE:? WARD:OUT DOCTOR:?

***

Output From Trial Analysis (1)

Fig 3-10.



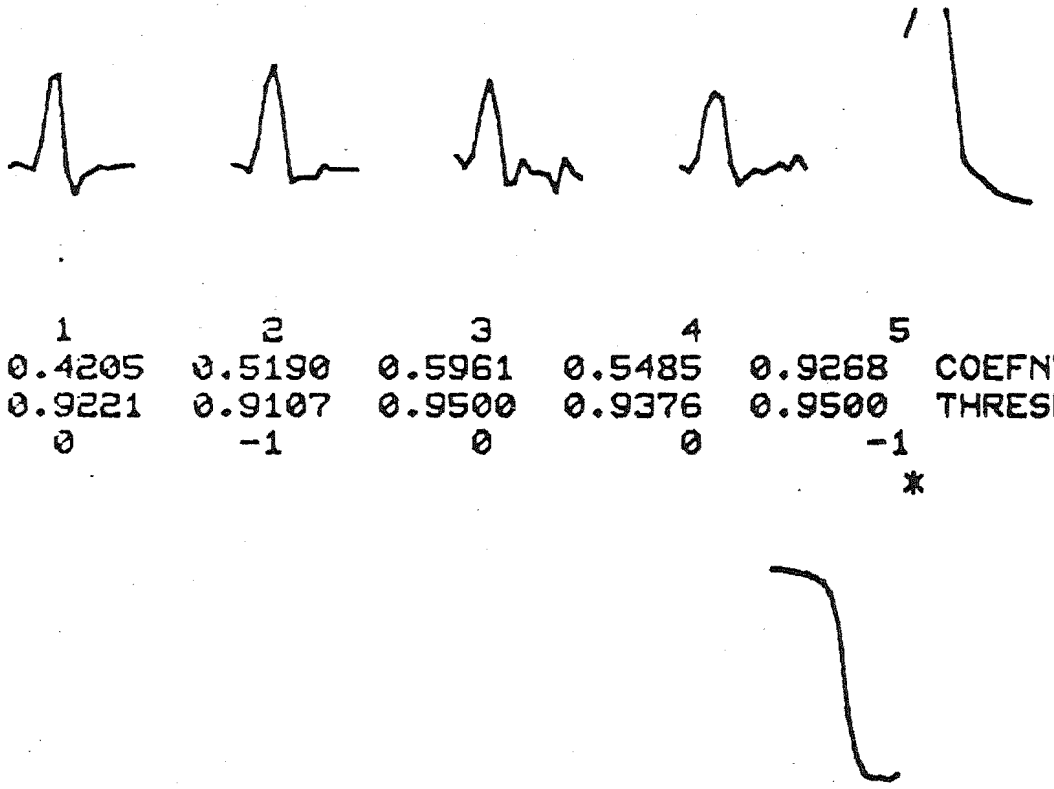
Section A.

NAME:T G. AGE:UNKNOWN WARD:OUTPATIENT DOCTOR:I

***

Output from Trial Analysis. (2)

Fig 3-11.



```
  1        2        3        4        5
0.4205   0.5190   0.5961   0.5485   0.9268   COEFNTS
0.9221   0.9107   0.9500   0.9376   0.9500   THRESHOLDS
  0        -1       0        0        -1         SHIFTS
                                      *

BEAT:  2623
TIME:12:34:27
RRINTERVAL=    0.6 SECS
RR PERCENTAGE=  73.2
Unmatched beat
ERR>
```

'Unknown' Beat Display to Operator

Fig 3-12.

Section B of figure 3-12 shows the beat that cannot be matched and section C gives the timing relationship of the unmatched beat with those beats immediately before it. A prompt to the operator to enter one of the possible options as specified in section 3.3.2 is also given.

CHAPTER 4

Performance of the Algorithm.

4.1 Digitization of the ECG Signal

Studies over the last 20 years have shown that an ECG waveform has spectral components from 0.05Hz to beyond 10KHz [8,9]. It is accepted, however, that a system with a pass band from 0.05Hz to 50Hz will show everything of clinical significance in the waveform [8]. This would suggest that a sampling rate of 100 times per second would suffice, and indeed several other automated analysis systems use this sampling rate [24,27,28].

However, there was some doubt as to whether the sampling rate was correct. since the normal QRS complex is approximatly 80 mSec wide and with a 100Hz sampling rate. This results in only 8 samples to characterise the position of interest in the waveform. To test if the 100 Hz sampling rate was sufficient to allow correct analysis by the algorithm proposed, a portion of an ECG recording, digitized at a 1KHz rate was analysed. The results obtained were similar to those for the 100Hz sampling (see table 4.1).

Beat number	Original file sampled at 1KHz	Four derived files, each taking every tenth sample from different starting points.			
1	20	4	2	2	3
2	22	2	2	3	2
3	22	3	2	2	2
4	11	2	1	1	1
5	54	4	5	6	4
6	9	1	2	1	1
7	24	3	2	2	2
8	29	2	2	3	2
9	20	2	2	2	1
10	27	3	2	3	3
11	42	3	4	5	4
12	26	3	2	3	2
13	25	2	3	3	3
14	28	3	3	3	3
15	23	3	2	3	2
16	22	2	2	2	2
17	26	2	2	3	3
18	21	2	2	2	2
19	32	3	3	4	3
20	28	3	2	3	3

Comparison of Accuracy of 1KHz vs. 100Hz Sampling

Table 4-1.

To verify that the instant at which the sample was taken had no significant effect on the resulting digitised waveform, a series of secondary files were produced from the 1Khz rate file, each containing every tenth sample but starting on a different sample. Figures 4-1 to 4-6 show the same beat from different starting samples and show that the basic features present in the 1KHz sampled waveform (fig 4-1) are present in all of the derived waveforms.

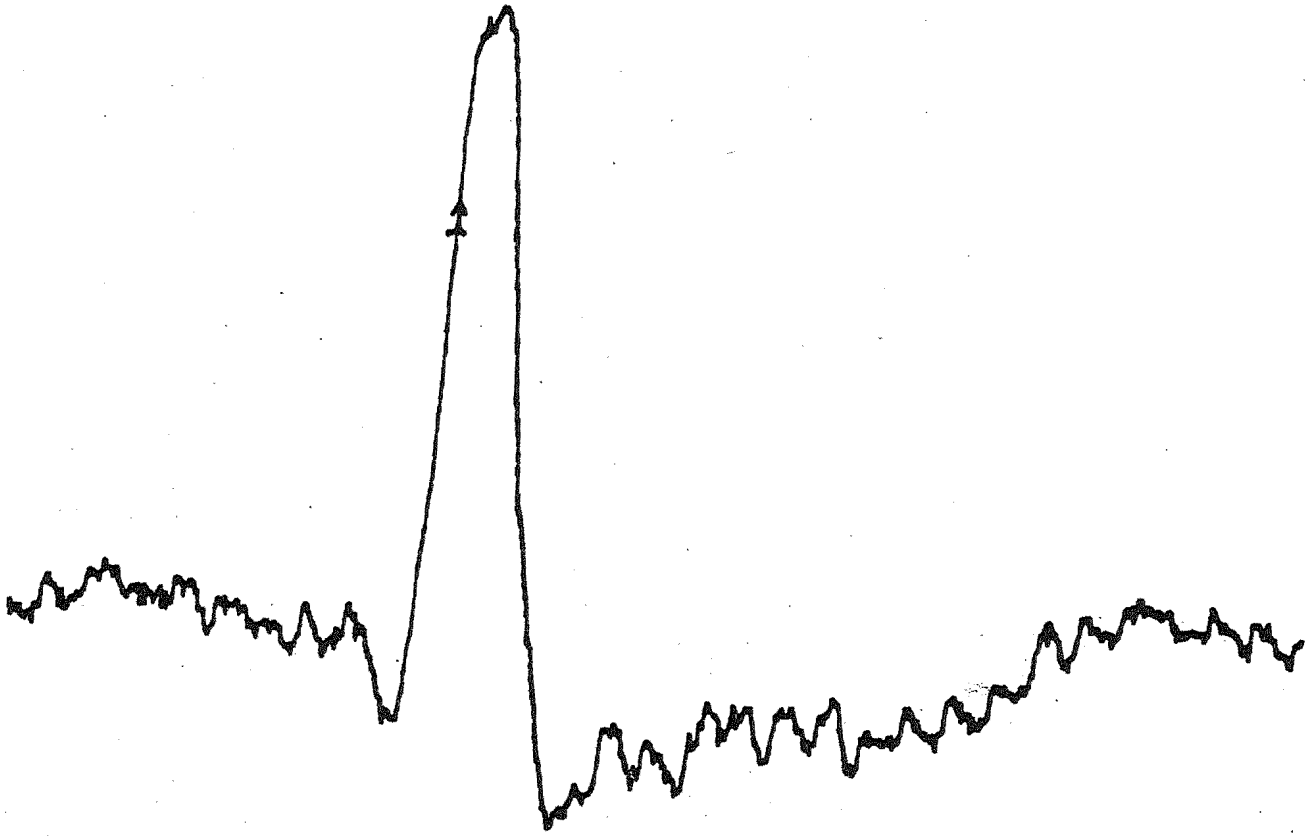
4.2 Evaluation of Algorithm

4.2.1 Method of Evaluation

In many cases it is difficult, if not impossible to accurately classify an ECG beat and it is common for experienced cardiologists to provide different diagnoses from the same beat. Consequently it is impossible for a computer program to supply answers that are 'correct' in everybody's opinion. Therefore the following evaluation method has been deliberately designed to test how well the system's results match those intended by the operator.

The adaptability of the algorithm to a variety of normal QRS shapes, to the particular abnormalities that each patient might have, and to the sensitivity of analysis by the operator have been mentioned as features of the algorithm. Unfortunately it is these features that make evaluation of the algorithm's performance difficult as the choices of templates and thresholds have a direct bearing on the algorithm's analysis of following beats.

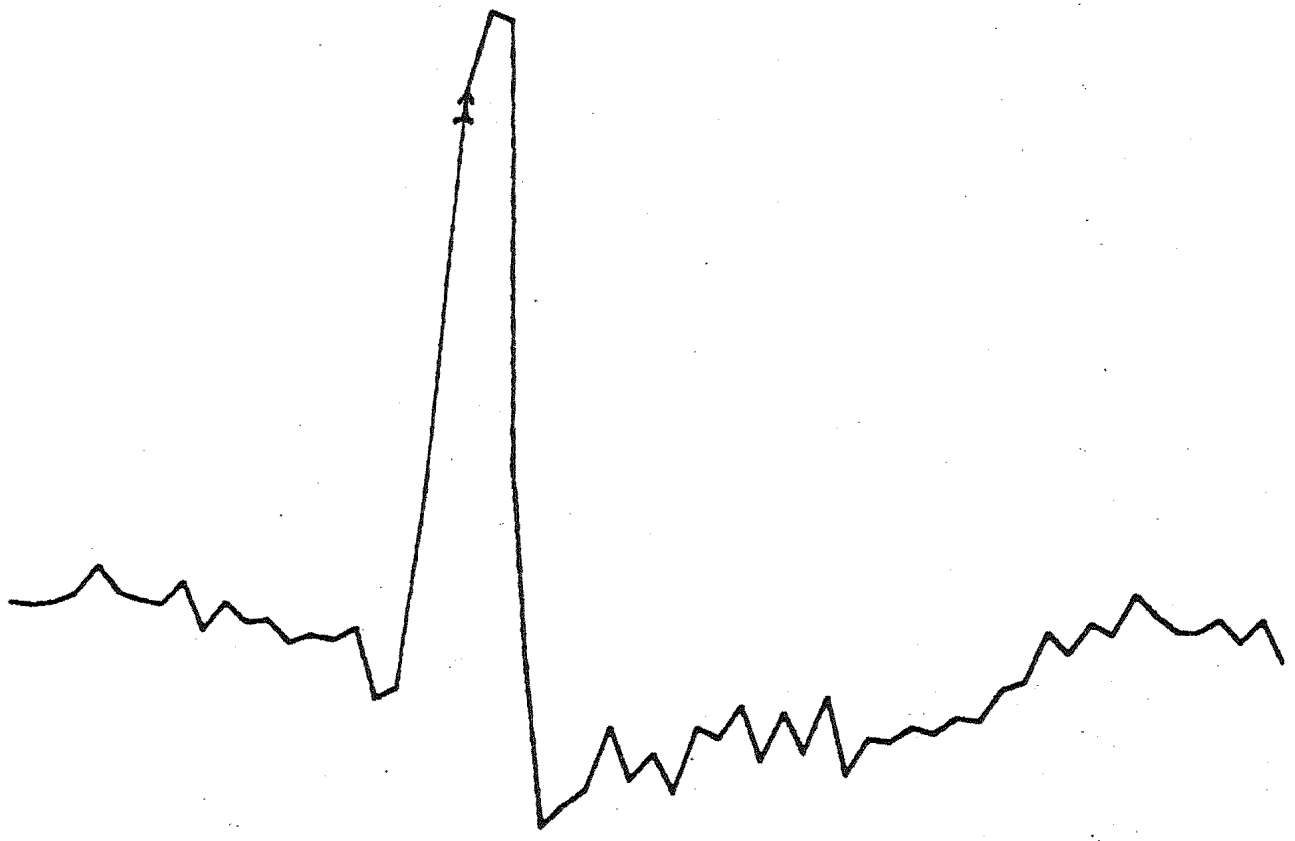
FILE ECGHAB BEAT 2
ECG>



Beat Sampled at 1KHz.

Fig 4-1.

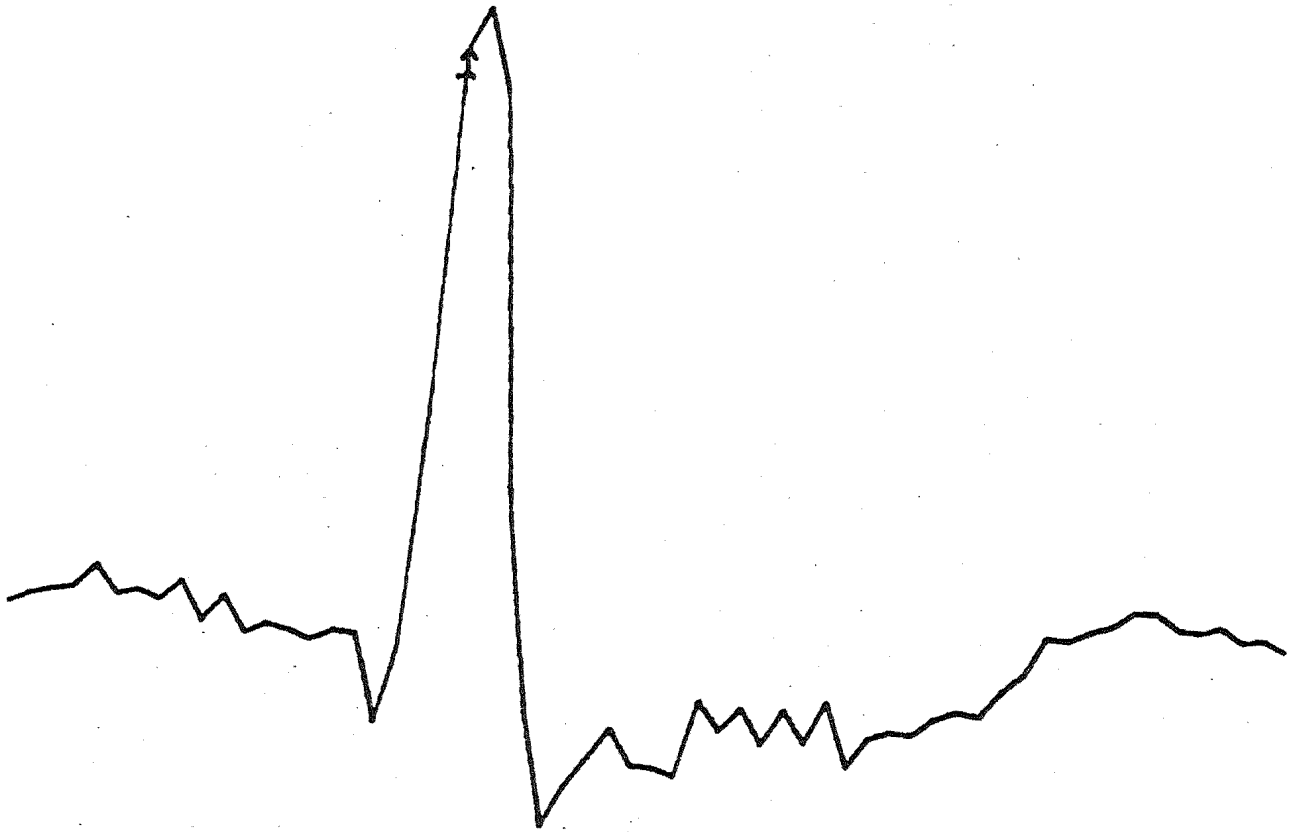
FILE ECGTB0 BEAT 2
ECG>



Derived Beat Sampled at 100 Hz (1)

Fig 4-2.

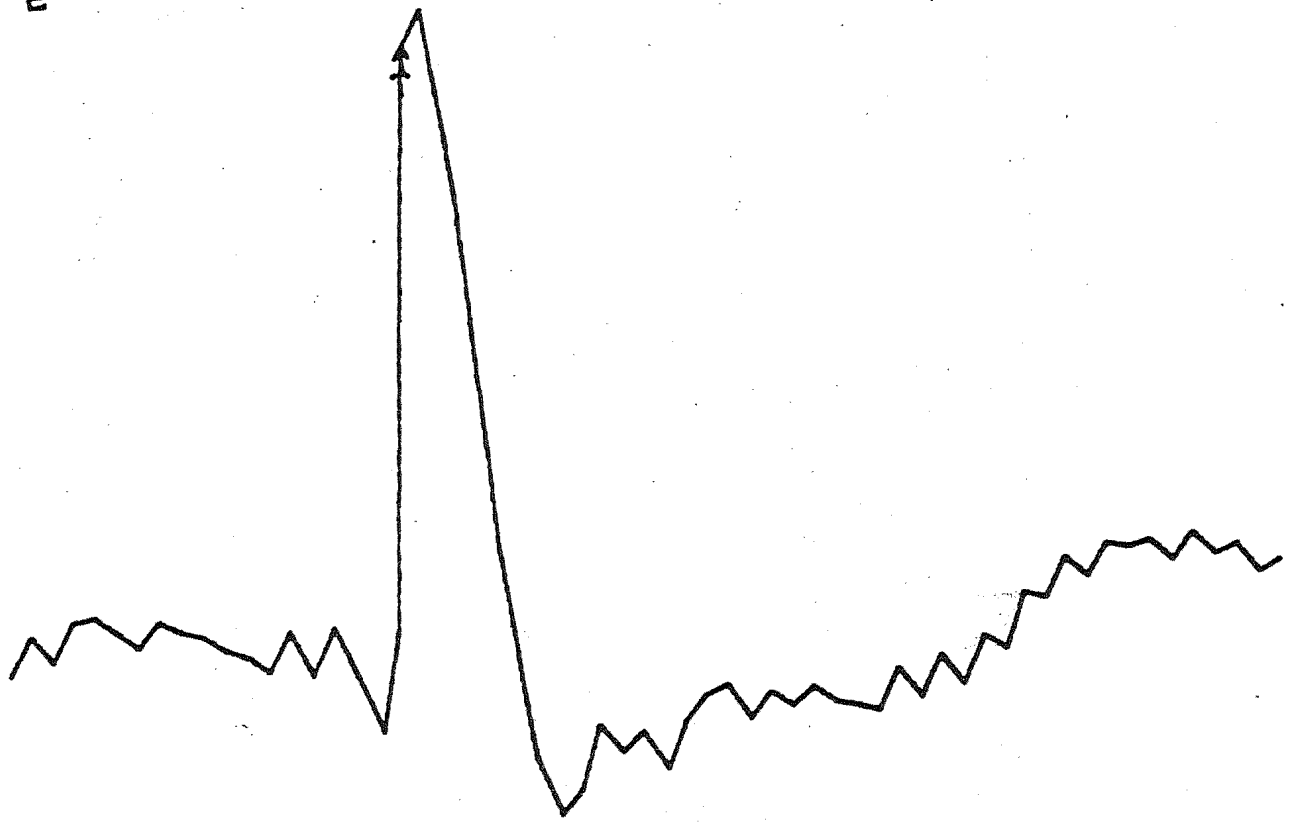
FILE ECGTB2 BEAT 2
ECG>



Derived Beat Sampled at 100 Hz (2)

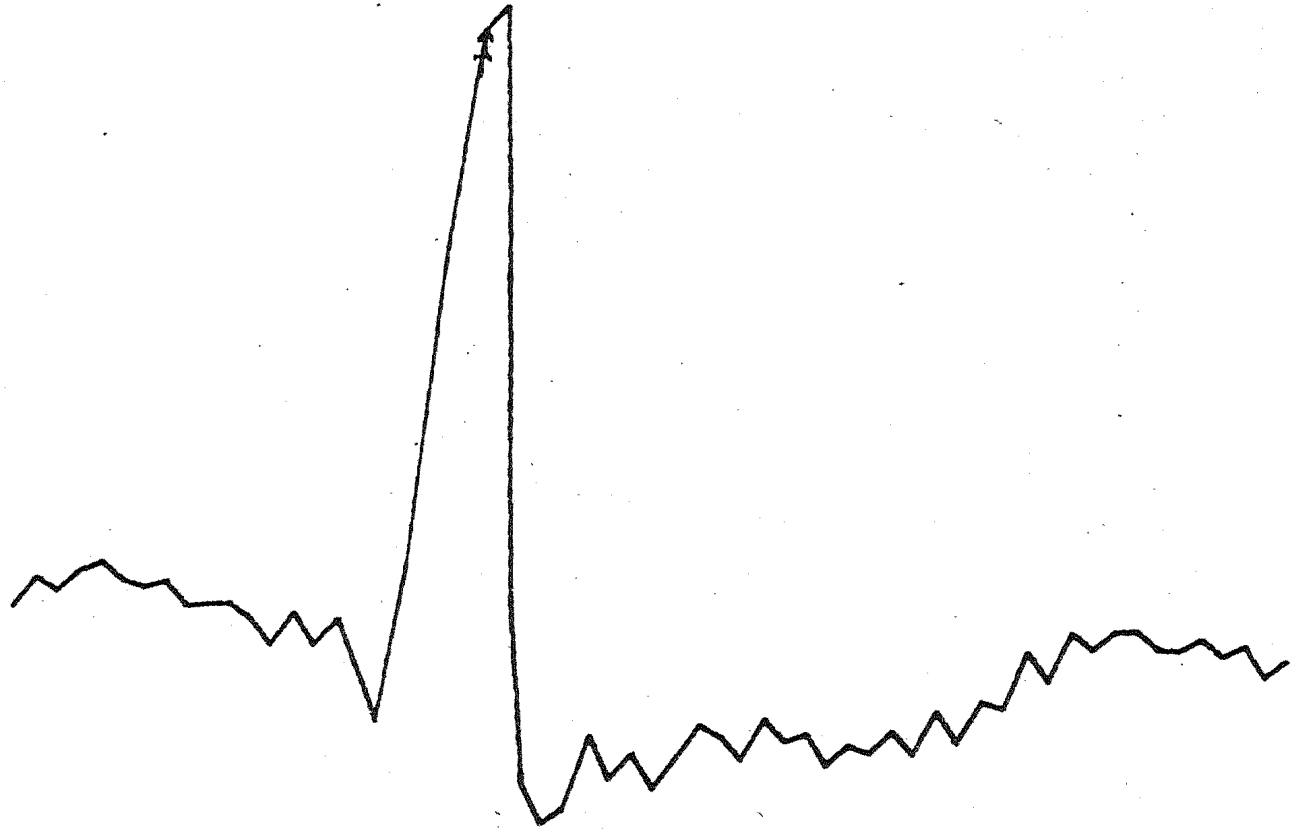
Fig 4-3.

FILE ECGTB4 BEAT 2
ECG)



Derived Beat Sampled at 100 Hz (3) Fig 4-4.

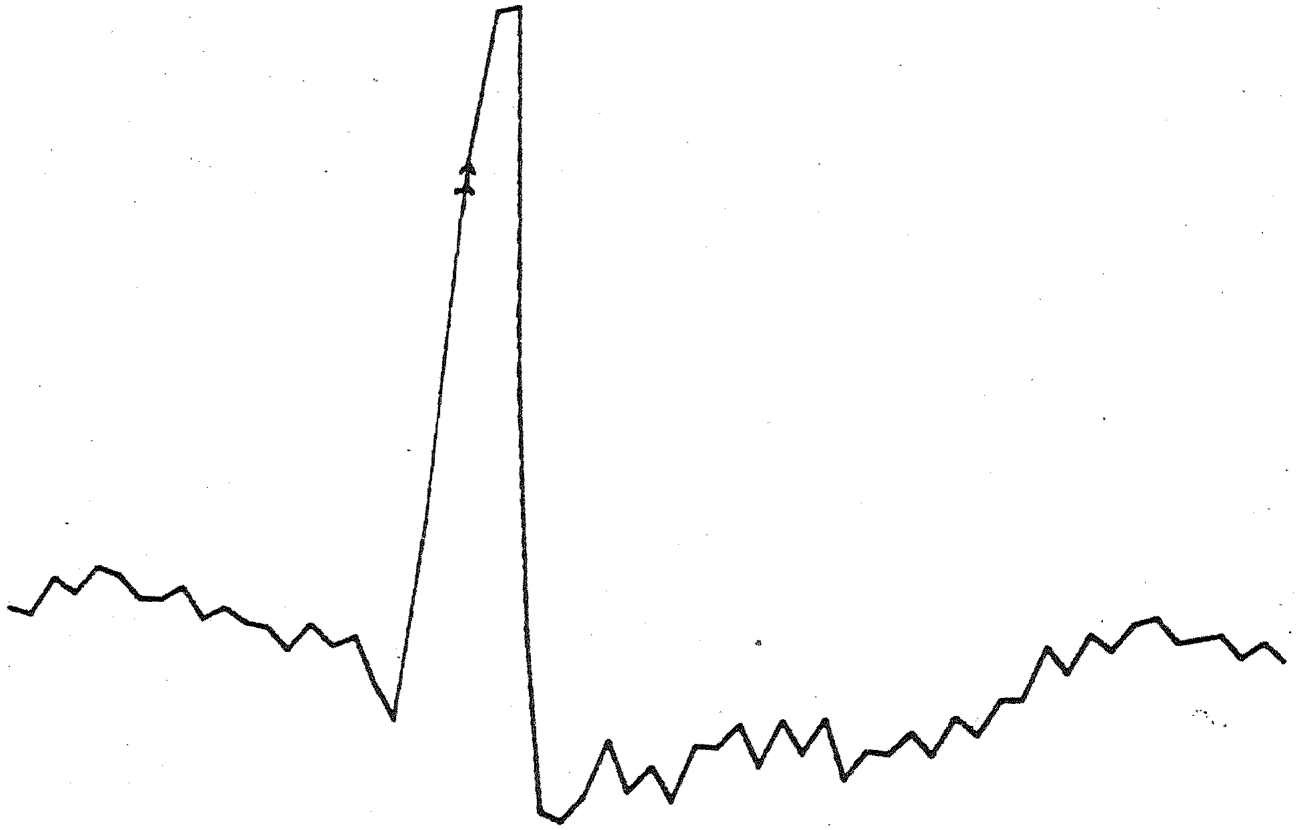
FILE ECGTB6 BEAT 2
ECG



Derived Beat Sampled at 100 Hz (4)

Fig 4-5.

FILE ECGTBS BEAT 2
ECG>



Derived Beat Sampled at 100 Hz (5)

Fig 4-6.

A series of five ECG tapes,recorded in the routine work of the hospital's cardiology unit, were selected and two or three five to ten minute (512 beat) segments were taken from each. The only criterion for selection of a segment was that there was to be some identifying feature within the recorded period. This was done so that each beat on the disk file could be compared with the same beat on a continuous strip chart recording of the segment.

Mr P. Bones, a bio-engineer in the Cardiology Department, manually inspected portions of the original tape recordings to find the approximate normal and abnormal wave shapes. He then performed the operator's function during the computer analysis of the recorded segments from that tape. This procedure approximates the actual analysis method envisaged for the final system, where the operator will be looking at the tape while setting the system up, and also to find a suitable 'normal' to give to the algorithm as its first template.

After each analysis, the beat by beat analysis of the results were stored on disk and the output plot of the run was compared with the results taken manually from a strip chart recording.

4.2.2 Results of Evaluation [30]

The results obtained from a beat by beat comparison of eight tape segments, analysed by both a cardiology bio-engineer and the computer algorithm, are given in table 4.2.

File Number	Normal Beats Number On Tape	Normal Beats Number Detected	Abnormal Beats Number On Tape	Abnormal Beats Number Detected	Number Of Missed Triggers	Number Of Beats Rejected As Noise	Comments
1	509	495	1	1	0	14	All beats were correctly analysed. R-R intervals down to 400 mSec timed correctly.
2	509	497	2	1	1	12	As the hardware trigger failed to detect the first abnormal beat, it was not analysed.
3	512	508	2	2	4	0	Contained VPB's that were detected by shape alone as timings were within the +/- 20% thresholds used in the R-R interval analysis.
4	569	509	0	0	59	1	Bigeminal rhythm with some R-R intervals less than 400 mSec and therefore ignored.
5	487	485	23	23	1	2	Erratic operation of the hardware trigger made some normal beats appear to have abnormal timings but normal shapes.
6	114	114	391	391	0	5	Contained a large number of beats generated by the patients demand pacemaker which were classed as abnormal. A third beat shape also present was correctly identified each time.
7	53	53	452	451	1	2	Same tape as file 6. One abnormal beat not analysed due to faulty triggering.
8	510	508	2	2	0	0	R-R interval thresholds masked several nearly abnormal timings.

Summary of Results of Algorithm's Analysis. Table 4-2.

Each segment analysed by the algorithm was exactly 512 beats long but the first two beats were only used to set up the initial timing parameters.

4.2.3 Performance of Algorithm

In drawing conclusions from the results in table 4.2 the distinction must be made between the operation of the algorithm and the operation of the system as a whole.

The hardware trigger failed in several instances to detect beats, and provided false triggers on some noise bursts. The former case lead to several abnormal beats not being detected at all and the following beat being classed as abnormally long. This is not a fault with the algorithm as it was not provided with correct information.

In the latter case above, help from the operator was requested each time a noise burst was found and a 'reject as noise' command was given. thus the noise burst was ignored and normal processing continued.

In both situations the algorithm performed correctly on the data it was given. These situations do, however, show the importance of reliable triggering for the proper analysis of a tape and the possible need for an automatic noise detection and rejection system to be included.

In one recorded segment, a bigeminal rhythm that was present caused the time between two beats to fall below the minimum of 400 mSec that the present implementation could handle. Again the algorithm correctly analysed the information it was given. However the R-R interval plot did show the missed beat as an abnormally long period to the following beat and it was flagged as such.

Thus, bearing in mind the shortcomings of the system around it, the algorithm correctly analysed the 4608 beats presented to it.

4.3 Suitability of Implementation Method

As the final system will be required to run competently at 60 times real speed, the limitations associated with the execution speed of the program will not be discussed here but in the hardware implementation proposal (section 5.2).

The performance of the algorithm is determined to a large extent by the information given to it by the operator. Thus the operator's interface and the experience of the operator play important parts in the proper running of the whole system. The technician who is most likely to be using the final system was asked to perform an analysis using the present computer implementation. Also two cardiologists were asked to comment on the format of the output plot.

From the operator's point of view the commands required by the implementation were found to be a little cumbersome and the lack of a

noise detection and automatic rejection system often resulted in unnecessary operator intervention. The information provided to the operator when the algorithm was requesting help was found to be basically good but the display of the unrecognised beat did not show enough of the waveform before and after the beat in question for the context of the beat to be properly determined. A display of at least one second of the ECG trace on either side of the beat in question, preferably more, was requested.

On the output plots that covered more than 2 hours, the 'Early/Late' and 'abnormal shape match' indicators were found to be of little use when large numbers of abnormalities were present. This was due to the resolution of the graphics terminal being too low to separate adjacent runs of abnormalities (fig 4-7). The abnormality indicators worked well for all time scales in identifying infrequent events, and provided time estimates to within 5 minutes of the position of the event on the tape.

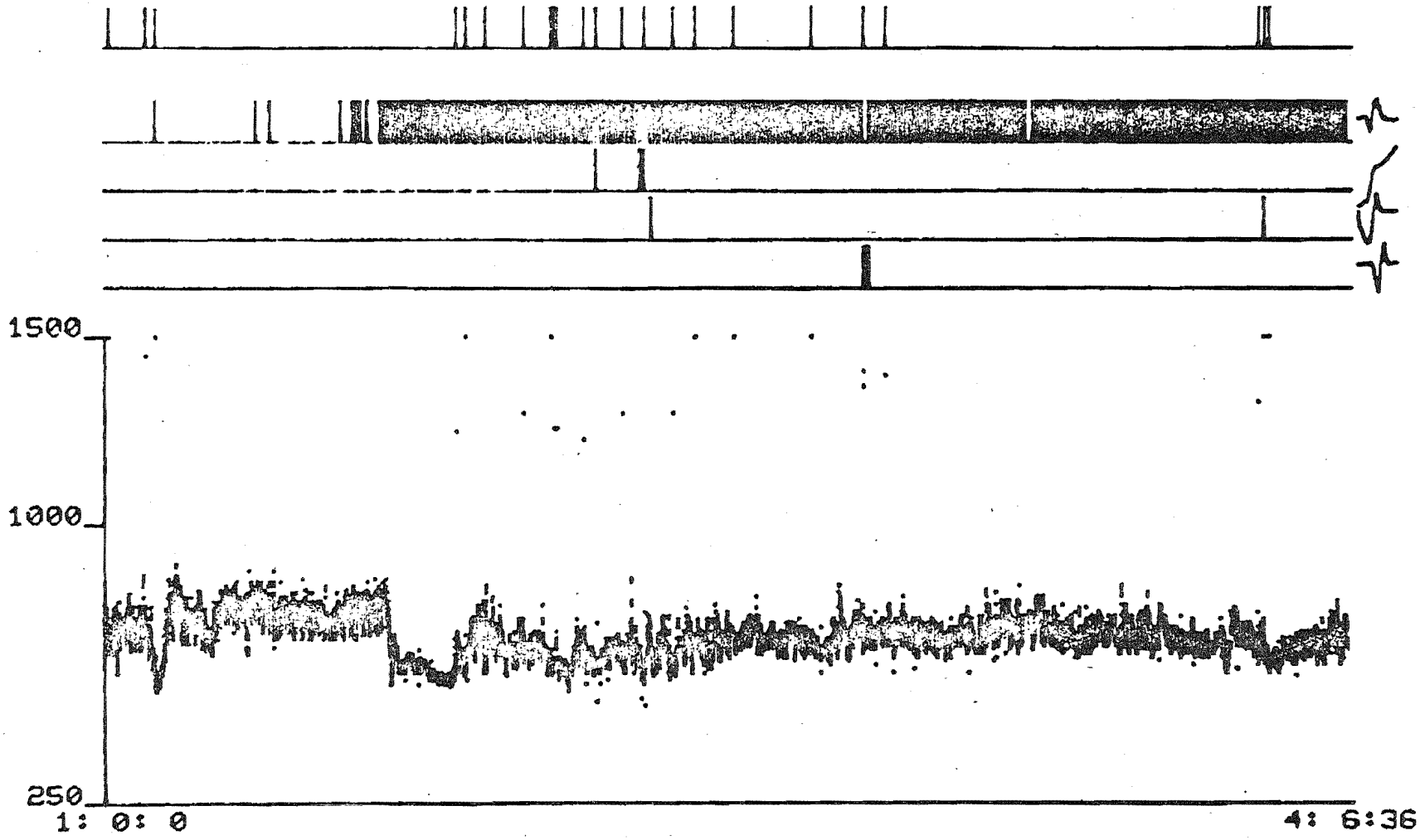
The cardiologists found the R-R interval plot very useful, but were initially confused by the abnormality indication section of the result output plot because this information is not normally available and they were not certain how to interpret it. The technicians found the whole output plot useful for identifying regions of interest.

The computer had only one graphics display device on it and so this was used to provide both the operators decision-making information and the display of the results of each beat's analysis but

PRINCESS MARGARET HOSPITAL CARDIOLOGY UNIT

BEATS IN FILE: 16000 STARTING TIME: 1: 0: 0 END TIME: 4: 6:19

STARTING DATE:14-Mar-80



NAME:T G AGE:UNKNOWN WARD:OUTPATIENT DOCTOR:

***

Low Time Resolution Output

Fig 4-7.

only one at a time, erasing one and drawing the other as required. This meant that the display was difficult to watch when frequent operator intervention was required. The problem would be overcome if two output devices could be used, one for the operator intervention and the other to plot the R-R interval and abnormality indications.

A comment was passed that noise recorded on the tape can affect the analysis system in two ways. The first occurs with random noise spikes causing the hardware trigger to indicate a false beat. Often the trigger will also pick up the next actual beat and thus the available command to completely ignore a noise induced trigger can correct this situation.

The second way in which noise can affect the analysis system is when noise is superimposed on a beat,causing the shape of the beat to be altered and also possibly causing the hardware trigger point to be slightly moved. In this situation the timing information available is still useful, even if the shape has been corrupted. It was therefore suggested that another command be introduced to reject the shape analysis results while retaining the timing information.

CHAPTER 5

Proposed Hardware Implimentation.

This thesis describes the development and trial implementation of an algorithm to analyse long term ECG recordings. The trial implementation, carried out on a PDP-11/10 minicomputer, made no attempt to work at the 60 times real speed that is required to keep up with the data being received from the tape replay unit. It is the intention of this section to outline the requirements for a hardware implementation of the algorithm that is able to work at the required speed (fig 5-1).

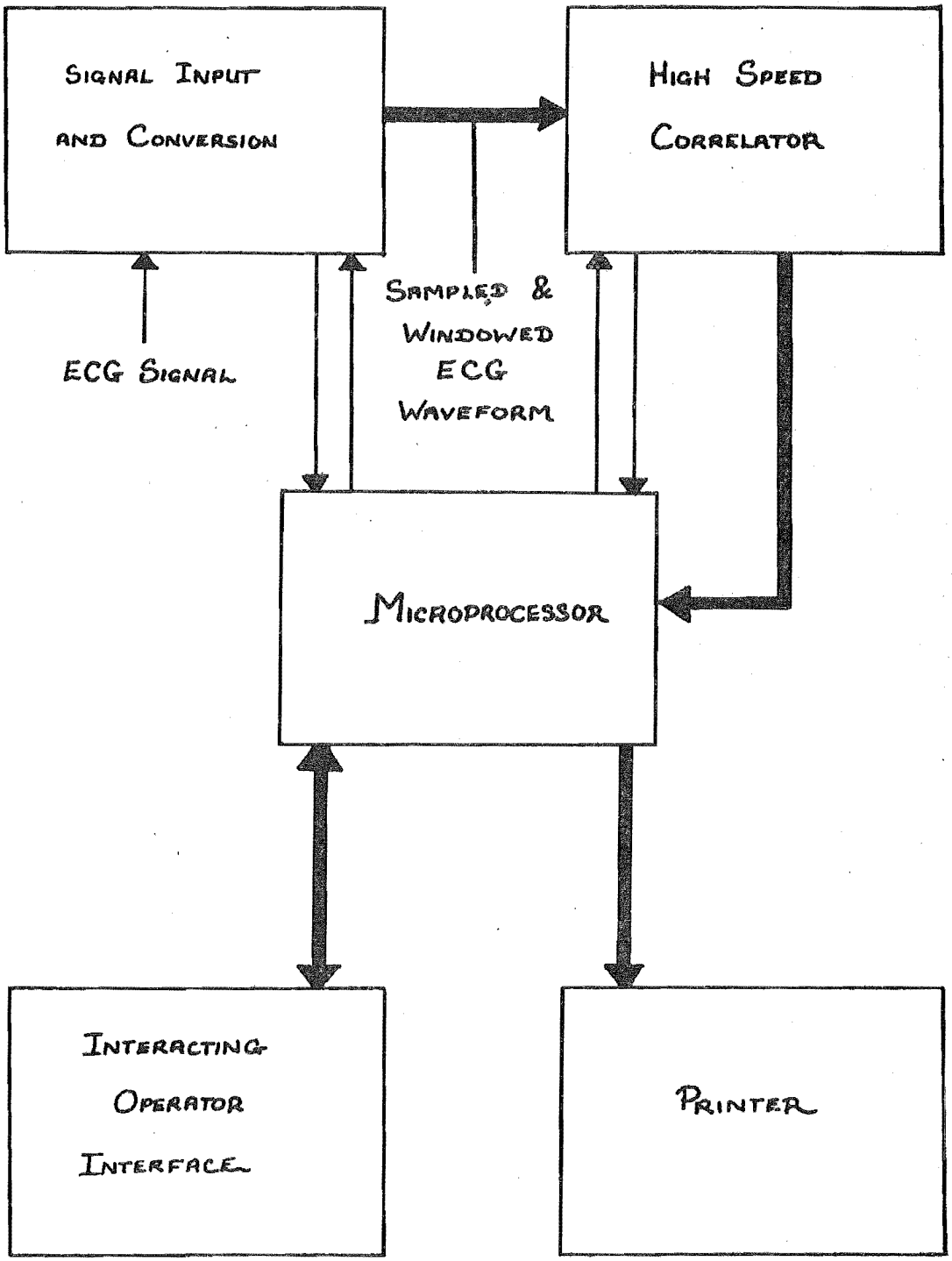
The total system separates into four regions:

1. Sampling and windowing the incoming beats

2. Correlating each beat with the current templates

3. Classifying each beat

4. Outputing the results of the classification



Block Diagram of Proposed Hardware System

Fig 5-1.

All These functions must be performed for each beat in under 6.7 mSec in order to handle heart rates of up to 150 beats per minute replayed at 60 times real speed.

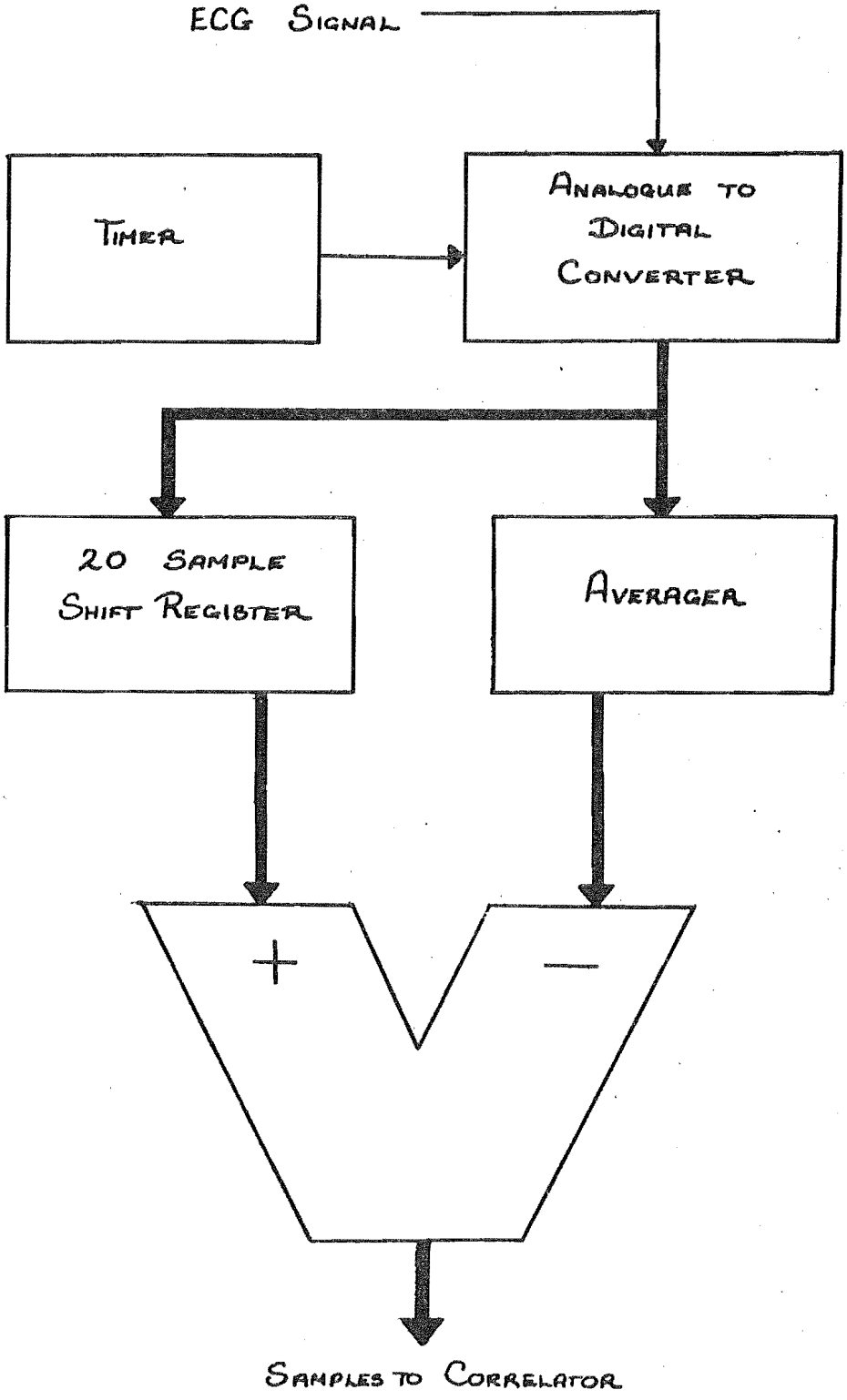
5.1 Beat Sampling and Windowing

A sample of the wave form is required every 167 uSec (sampling at 100 times/second at 60 times normal speed = 6000 samples/second). As illustrated in fig 5-2 a timer initiates an analog to digital (8 bit) conversion every 167 uSec. At the end of the conversion time the digital sample is entered into a 20 sample long shift register and also into an averaging circuit.The shift register holds the last 20 samples obtained, ready for the correlation process that starts when a complete beat has been obtained.

The averaging circuit maintains a value that is the average of all the samples in the shift register at a particular time. When the correlator process requests the stored samples, the average value is subtracted from each sample before it is passed on, as is required by the correlation formula used (see section 3.3.2).

This section of hardware also converts the trigger signal into a 'beat ready' signal for the following sections.

5.2 Correlation Function Implementation and Beat Classification



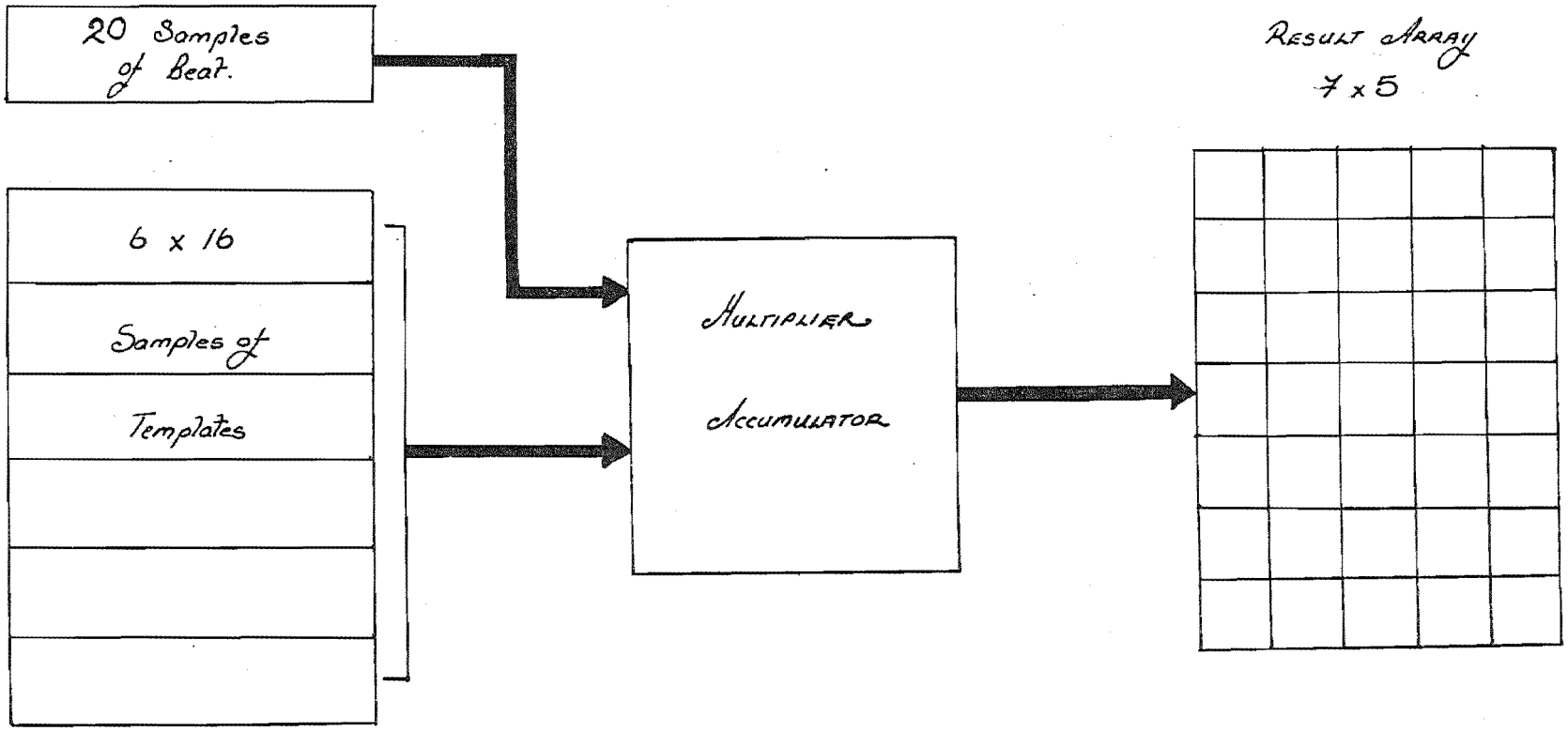
Signal Input & Conversion

Fig 5-2

The calculation of the correlation coefficients (see section 3.3.2) for a beat versus a template required 16 multiplications and additions for the dividend and 17 multiplications, 15 additions and one square root for the divisor. Experience with the mini-computer trial implementation has shown that one normal and five abnormal templates are sufficient to cover most analyses. However this leads to a significant number of arithmetic operations (390) to be performed within the 6.7 mSec available or one operation every 17 uSec without any other processing being done. The proposed solution to this problem is outlined below (fig 5-3).

The major control element of the whole system is a microprocessor. The processor will take samples from the input A/D converter and place them into special purpose storage registers that are accessible to a hardware correlator (see below). The correlator will be instructed to perform the calculations required for the divisor of the correlation coefficient equation of the beat with each defined template, and also to produce an auto-correlation value.

The hardware correlator looks to the storage registers where the processor has stored the samples of the template and the beat. Corrresponding samples from the beat and each template in turn are multiplied and the sum of products for each template is stored in another set of storage registers, ready for the processor to use. This process is performed five times, once for each shift of the incoming beat (see section 3.3.2). The auto-correlation is performed by using the beat itself as the template.



High Speed Correlator

Fig 5-3.

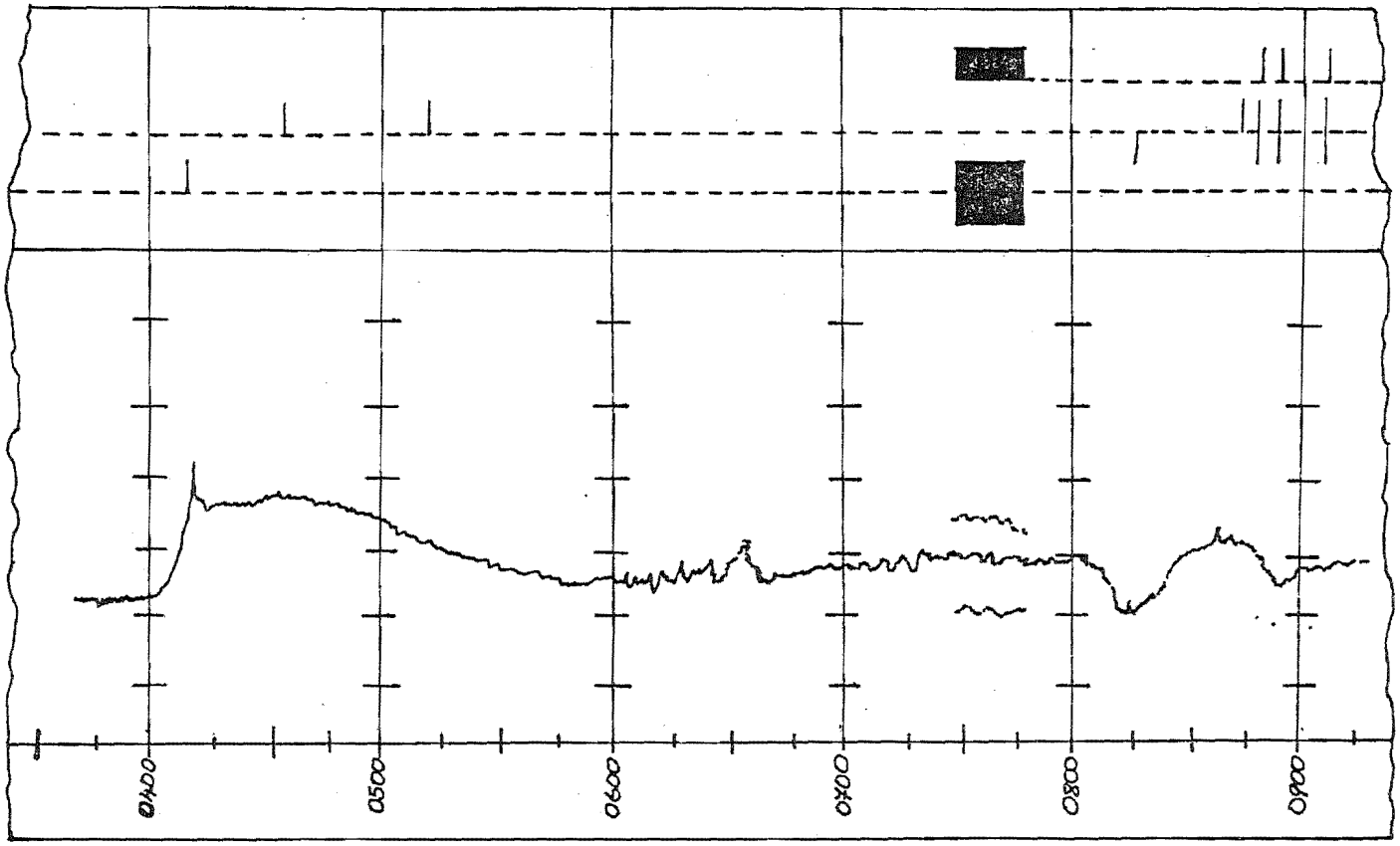
The processor is thus provided with the dividend and the beat normalisation constant for each template. The template normalisation constant is retrieved from memory and the processor completes the coefficient calculations and decision making process (see section 3.3.2) with the aid of a hardware multiply/divide integrated circuit. The square root can be avoided by remembering the sign and then squaring the dividend, the result then being the square of the correlation coefficient.

The processor is then able to retrieve the timing information from the input section and produce a final decision on the beats classification as per section 3.3.3.

5.3 Output of Results.

The results from the processor's decision making tell which group the beat belongs to and whether it is early or late. This, together with the R-R interval for the beat, must be output to a hard copy device. A format similar to that used in the trial implementation is proposed (fig 5-4).

The output device must be capable of high speed graphics with high resolution as it is expected to plot the information about a beat in under 6.7 mSec. Because trend information is all that is required, this restriction can be relaxed slightly in that the data derived from several (say 10) beats can be output on a single line for high speed



Abnormal Shaped Beat markers

Early / Late markers

R-R interval Plot

Time Markers

Proposed Analysis Output Format

Fig 5-4.

plotting. Also if a suitable dot matrix printer is used then approximatly 7 lines can be plotted simultaneously, thus reducing the print requirements to one printer line every second for high time resolution of the analysis results.

It is therefore proposed that a dot matrix printer, with the capability of addressing individual dot positions, be used to output the results.

5.4 Operator Interaction Terminal

Before requesting help from an operator, the system must supply information as to the shape of the current templates and the unmatched beat. Experience has shown that this display should show as much of the wave shape on either side of the beat as possible, certainly not less than +/-1 second. This is to allow the operator to put the beat into its proper context before making a desicion.

In view of this and the desire not to break up the permanent output of results with information only for the operator, a device separate from the print out device above is required. The second output device needs to be interactive and erasable and thus a CRT terminal with graphics capability is recommended. If cost limitations dictate, a simpler terminal of LED displays with a special purpose keyboard could be used in the meantime.

A hard copy device, such as an X-Y plotter, is not recommended as there may be a large number of requests to the operator, especially at the start of an analysis while the templates and their thresholds are being established.

CHAPTER 6

Discussion.

6.1 Conclusions

There is a need for rapid and accurate analysis of long term (24 hour) ECG recordings in order to highlight regions of abnormal activity. Other groups of workers throughout the world have recognised this need and have produced working systems that in the main use dedicated mini-computers or large time-shared computer systems and produce a clinical classification of each beat detected. They all also produce trend information on heart rate, ectopic beats and other selected abnormalities at regular intervals.

This approach was considered inappropriate for the environment at Princess Margaret Hospital's Cardiology Unit where a small mini-computer was available at irregular intervals and it was impractical to purchase a mini-computer solely for this task.

People in general, including medically trained staff, tend to accept computers as infallible. Therefore it was considered unwise for the computer program to provide clinically oriented results direct

to medical staff as they might not always treat the results as a guide but rather as an absolute diagnosis.

A further consideration was the manual analysis method presently used in the cardiology unit. A technician visually scanned the recorded ECG, looking for abnormal rhythms and beats. Representative samples of these abnormalities were recorded on a strip chart and later presented to the doctor for clinical evaluation. Thus it was desirable to develop a system that was comparable with the present method with a minimum of effort by the medical staff.

The requirement was to develop an algorithm that would summarise a 24 hour ECG recording and show those regions that contained abnormal rhythms and beats for later detailed manual analysis. Such an algorithm must also be capable of being implemented by dedicated microprocessor based hardware.

The first attempt at developing such an algorithm was based on the fact that the major abnormalities to be looked for have characteristic timing patterns. A trial computer program soon showed that it was very difficult to accurately separate such abnormalities from normal variations in timing. Also there are significant abnormalities which show no such timing variations and so a form of shape analysis was sought.

Two of the more clinically significant features of an ECG signal are the width and height of the QRS complex. It was found that a

computer program written to determine these features could find the QRS height adequately but that the width calculation showed unacceptable variations.

Further investigations into shape analysis methods indicated that calculating the correlation coefficient between a QRS complex and a stored template was the best and most reliable method available. When used in conjunction with a timing analysis of preceeding beats this method could satisfactorally characterise any particular QRS complex.

The shape of a recorded QRS complex varies from one patient to another and is also dependent on electrode palcement. Thus it is not practical to have a set of standard templates for use on all patients and so a method of using the recorded beats themselves as the templates was developed.

A trial implementation of the algorithm has been made to test and to evaluate its performance on real but known data. This implementation has shown that the principle of the algorithm works well.

The tape replay unit supplies a signal to the algorithm to indicate when a QRS complex has been detected and this signal is also used for the timing analysis. Use of this signal was found to be satisfactory except when the recording was affected by large amounts of noise.

The developed algorithm required interaction with an operator for situations when a stored template could not be found that properly matched a particular beat. The scheme used to interact with the operator was found to be good, with the exception of not properly showing the context of an unrecognised beat shape. However the solution, to show at least one second of the ECG signal on either side of the QRS complex, is straight forward.

The form of the output for the results of the algorithm has been found to be acceptable to the medical staff and useful once the meaning of the extra information that is not usually available to them is understood.

The developed algorithm has thus been found to be satisfactory and has formaed the basis for a hardware system now under construction.

6.2 Suggestions for Furthur Work and Extentions.

A hardware implementation of the algorithm developed here is currently under construction as the principle of the method is considered to be satisfactory. The use of a microprocessor as the main control element and to perform the final classification still allows sufficient flexibility in the system to extend its capabilities, thereby improving its usefulness.

The present algorithm requires operator intervention in the case of an unrecognised beat being detected. If there is a significant amount of noise recorded with the ECG signal this can lead to a large amount of interaction being required which stops the operator from performing other duties and thus defeats one of the main features of the system.

Experience with the current tape replay equipment has shown that a major source of noise is the electrical activity of the muscles in the rib cage adjacent to the electrode sites. As attempts are already made to place the electrodes in areas that produce as little of this type of noise as possible, an attempt should be made to detect when such noise is present and to automatically reject any detected beats in this period.

The special purpose hardware under construction is designed to take a lot of the computing load off the microprocessor and to allow the system as a whole to work at 60 times real speed. However the low time requirements on the processor could also be used to advantage in a multi-purpose computer system that simultaneously monitors other physiological signals, such as respiration and blood pressure, from patients in real time.

While the present system has been designed specifically for monitoring ECG signals, it is quite possible for it to study (with appropriate modifications where necessary) any repetetive waveform which has unusual patterns occuring at unpredictable times. One such

waveform that is receiving more attention recently is arterial blood pressure. With the introduction and probable widespread use of multi-channel portable recorders, the simultaneous measurement of the ECG and arterial blood pressures is possible and so an extension of the algorithm into this field should be made.

APPENDIX A

Common Arrythmias.

1. Sinus Arrythmia.

As the oxygen level in the blood falls, one of the two interconnected nervous systems of the human body, the 'Parasympathetic', becomes more active in order to stimulate the chest and diaphragm muscles to initiate a breath. Conversly, as the oxygen level in the body increases the parasympathetic nervous system becomes less active.

However the heart is also connected to the same nervous system and thus also responds to the oxygen demands of the body; speeding up as the level falls and slowing down as the level increases.

The amount of variation in heart rate thus caused by respiration cannot be reliably predicted as it very much depends on the condition of a person's nervous system, but it can be as much as 20% of the normal heart rate.

Rather than being an abnormal rhythm, Sinus Arrythmia, as this type of heart rate change is called, shows that the heart is responding to the demands of the body and that the heart beats are

being initiated from the proper part of the heart.

2 Tachycardia and Bradycardia

Tachycardia is defined as three or more consecutive beats from the same origin at a rate exceeding 100 beats per minute [32]. There is an implication that the origin of the beats is not the Sinus Node where the beats should originate, but this is not necessarily so.

Bradycardia is similarly defined as three or more consecutive beats from the same origin at a rate below 40 beats per minute [32].

Both tachycardia and bradycardia are, in practice, not so well defined and are usually used as terms to describe a heart rate that is greater than, or less than, the usual rate of the patient and is not caused by a physiological need such as excercise.

3 Atrial and Ventricular Premature Beats

Both of these abnormalities have the same basic feature that the beat does not originate at the Sinus Node and occurs before the next sinus beat.

Atrial Premature beats (APBs) originate in the atrium between the Sinus and the Atrioventricular Nodes and thus appear to the ventricles to be a normal beat producing a normally shaped QRS complex. The stimulus also spreads back through the atrium and resets the Sinus

Node so that the next beat started by the Sinus Node will have a 'normal' R-R interval (ie there is a phase shift in the heart beat rhythm). Thus the characteristic of an APB is a normal QRS complex that is earlier than expected and resets the following 'normal' beat timing.

Ventricular Premature Beats (VPBs) originate in the ventricles and so generate an abnormal QRS complex due to the different conduction path. These abnormal beats are usually stopped from entering the atria as the stimulus meets the normal stimulus from the atria and the two stimuli cancel eachother out. Therefore the atrial pacemaker is not reset and the next beat occurs within the 'normal' pattern of heart beats.

APPENDIX B

Phase Alignment Technique.

Many analysis techniques require accurate knowledge of the position of the QRS complex within a segment of the ECG waveform. This phase alignment technique [10] is an attempt to locate a possible point of symmetry of a QRS complex that is repeatable for similar but not identical waveshapes, thereby providing an accurate reference point for further study of the QRS complex.

The output of the Fast Fourier Transform (FFT) is the magnitude and phase of a series of harmonics of the fundamental frequency defined by the width of a sampled window. The number of harmonics is determined by the number of samples in the window.

The magnitude information gives the size of each harmonic required to make up the sampled waveform, and the phase information shows the time relationship between the harmonics.

For this discussion it is assumed that an ECG-like waveform has been windowed and sampled so as to satify the Nyquest Criterion, which means in this case that the highest harmonic corresponds to approximatly 50Hz.

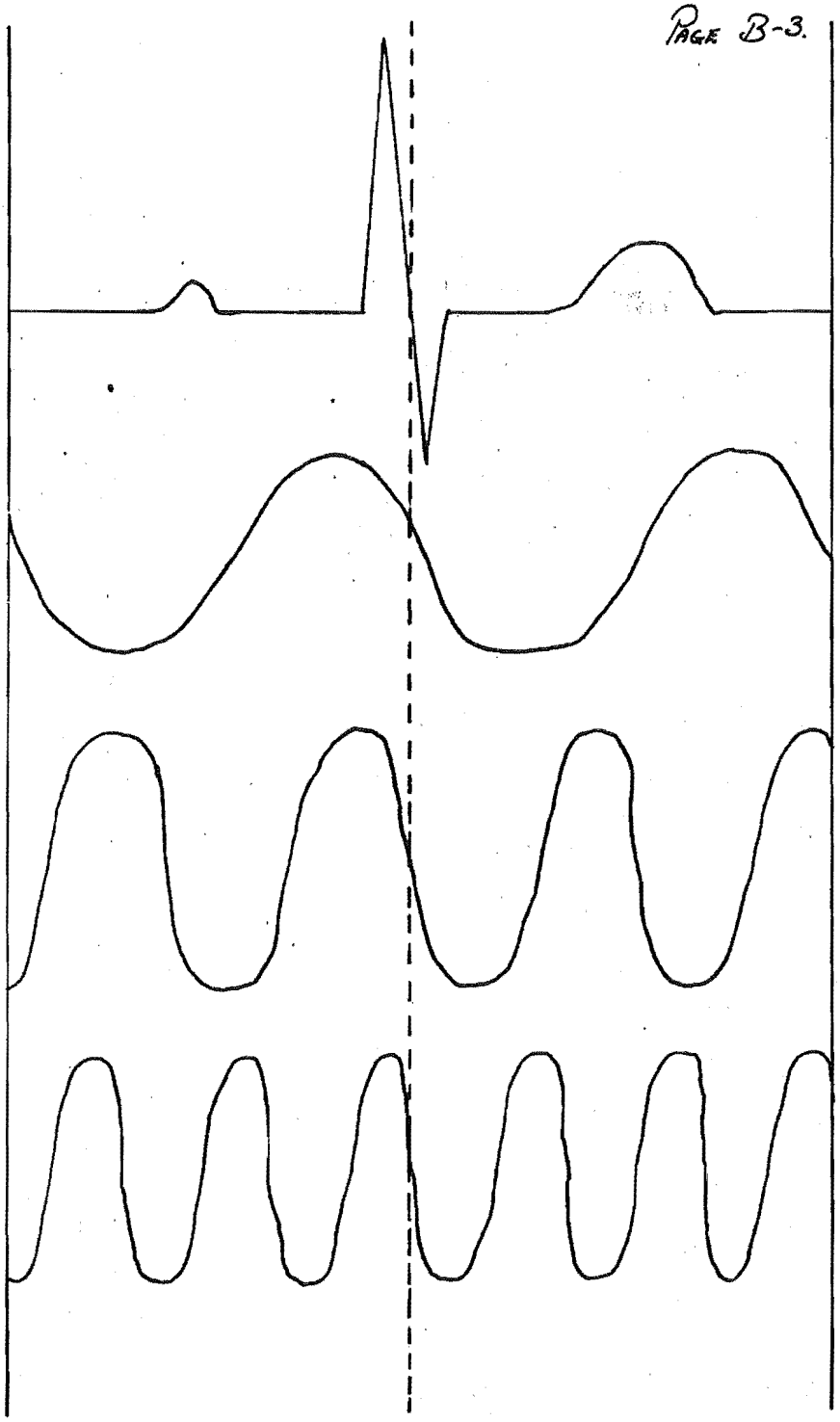
Such a wave form has one major feature, namely the QRS complex, which contains the majority of the high frequency information. Therefore there is a tendancy for the phases of the higher order harmonics to align themselves such that they all coincide near the axis of symmetry of the major feature (fig B-1).

The phase angle given by the FFT is always in the range 0 to 2π radians but this can have added to it up to $2(n-1)\pi$ where 'n' is the order of a particular harmonic (ie the fifth harmonic can range between 0 and 8π radians). It should be noted that when phase alignment occurs, the actual phase angle of the (n+1)th harmonic measured from some reference time will be greater than that of the nth harmonic (fig B-2).

The low order harmonics contain information about the QRS complex and the surrounding wave shape, and only the higher order harmonics contain information about the QRS alone. Therefore the 5th to 25th harmonics have been used in this application.

The 5th harmonic's phase angle is taken as a base value. The phase angle of the 6th harmonic is compared to that of the 5th and if less than the 5th has 2π added to it. The process continues until the last (ie the 25th) harmonic has been 'aligned'.

A least squares line is then fitted to the aligned phases and the 'y' intercept set to pass through the origin. The value of the resulting expression evaluated for the fundamental frequency gives a



Phase Alignment of Higher Order Fourier Harmonics on QRS Complex

Fig B-1.

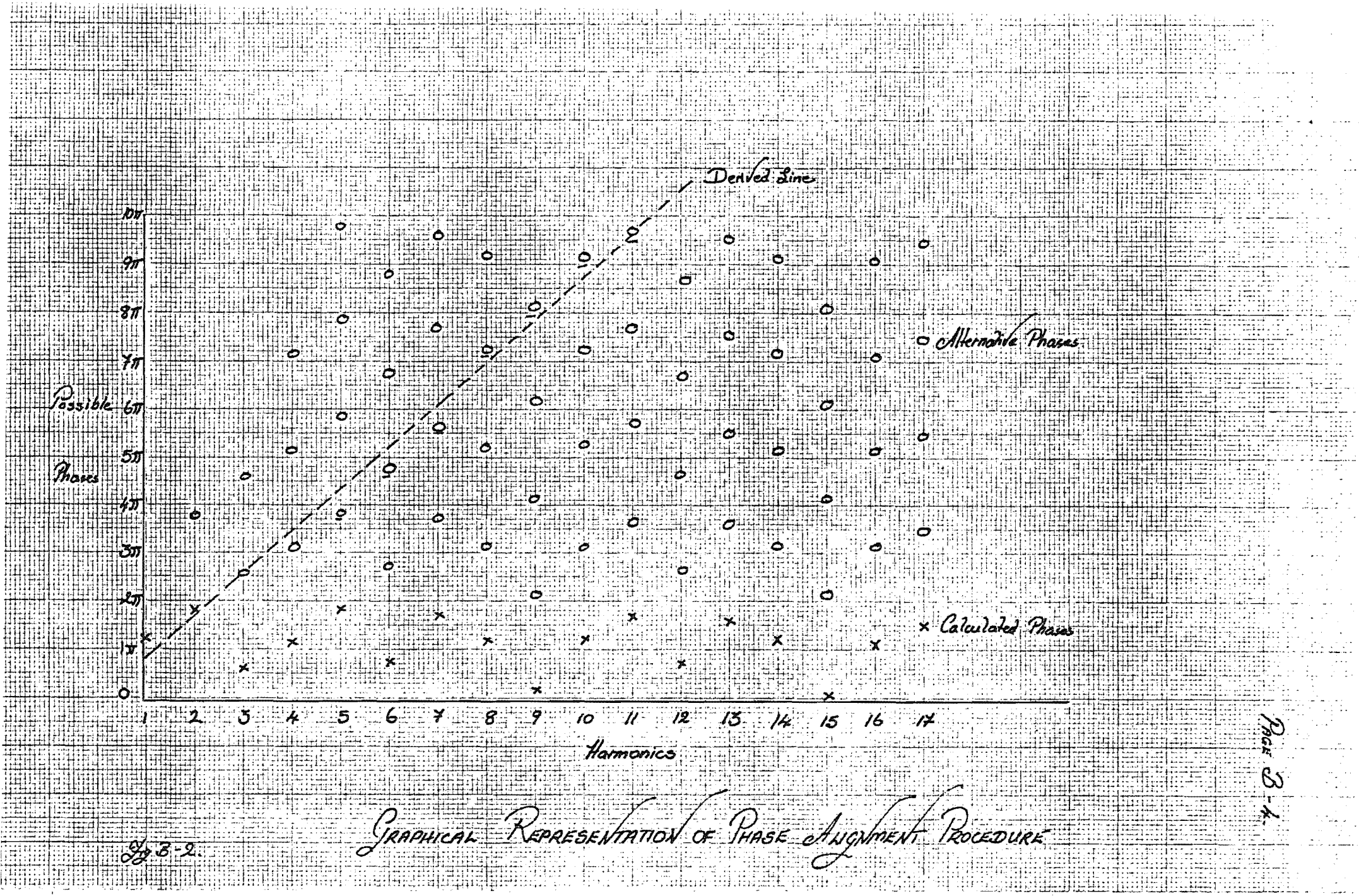


Fig 3-2.

GRAPHICAL REPRESENTATION OF PHASE ALIGNMENT PROCEDURE

phase between 0 and 2π which corresponds to the point of phase alignment on the fundamental, and thus uniquely defines the required point within the window.

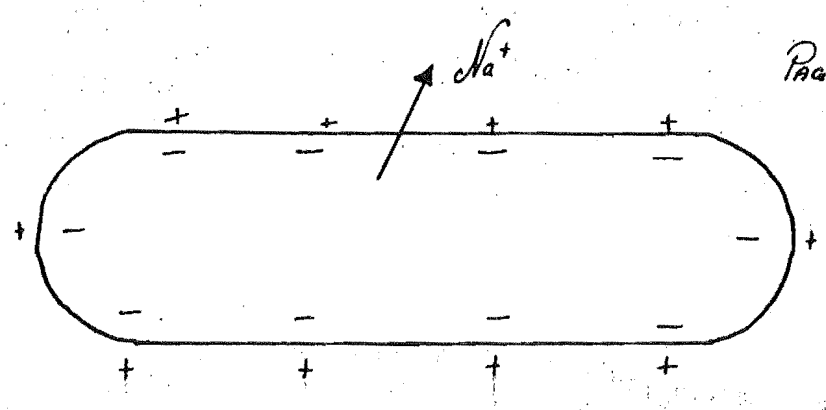
APPENDIX C

Physiology of a Muscle Cell.

When a muscle cell is relaxed, a chemical pump is expelling sodium ions from within the cell and so the inside of the cell has a net negative charge with respect to it's surroundings (fig C-1). When the cell is stimulated, the characteristics of the cell membrane alter slightly which results in an influx of sodium ions into the cell, resulting in a reversal of the charge accross the cell wall. This action coincides with the contraction of the muscle cell (fig C-2). Due to the length of a muscle cell, this depolarisation, as it is called, does not occur simultaneously throughout the cell but spreads out like a wave from the point of stimulation. Thus the outside of the muscle cell appears to be a dipole source while the cell is depolarizing.

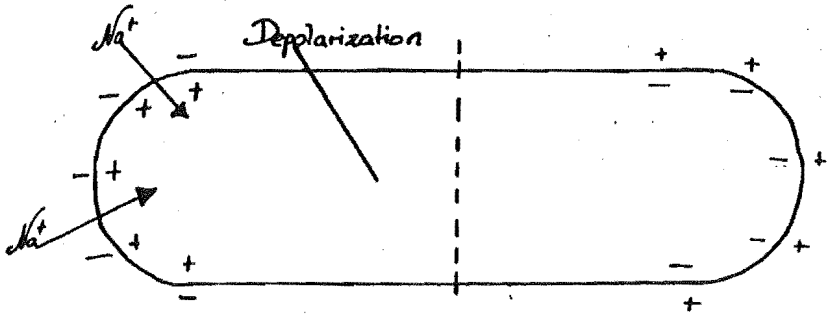
When the wave of depolarization travelling through one cell moves past a point of contact with another muscle cell, the second cell is stimulated and so the stimulus and resulting contraction of the muscle cells spreads through the whole muscle.

While the cell is depolarised, it cannot be restimulated and so a cell cannot reactivate itself, nor can it be restimulated by an



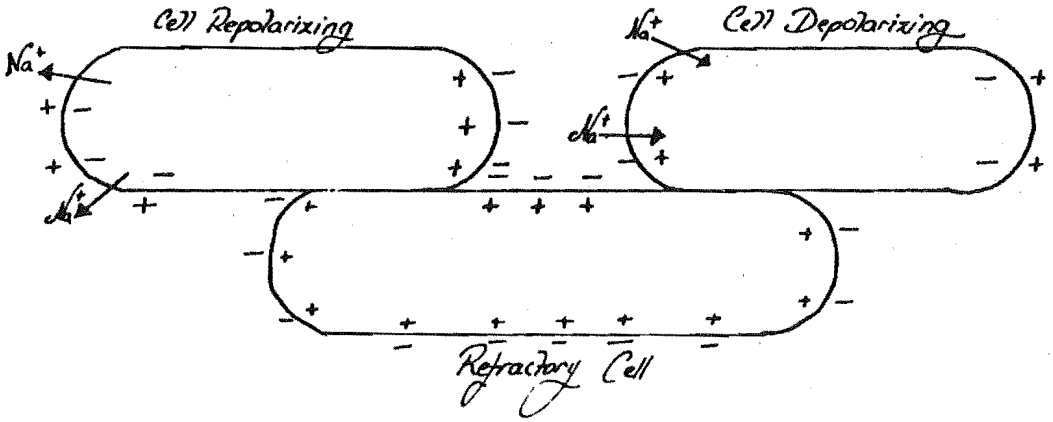
Effect of Sodium Pump on Muscle Cell

Fig C-1.



Depolarizing Muscle Cell

Fig C-2



Contraction Travelling One Way

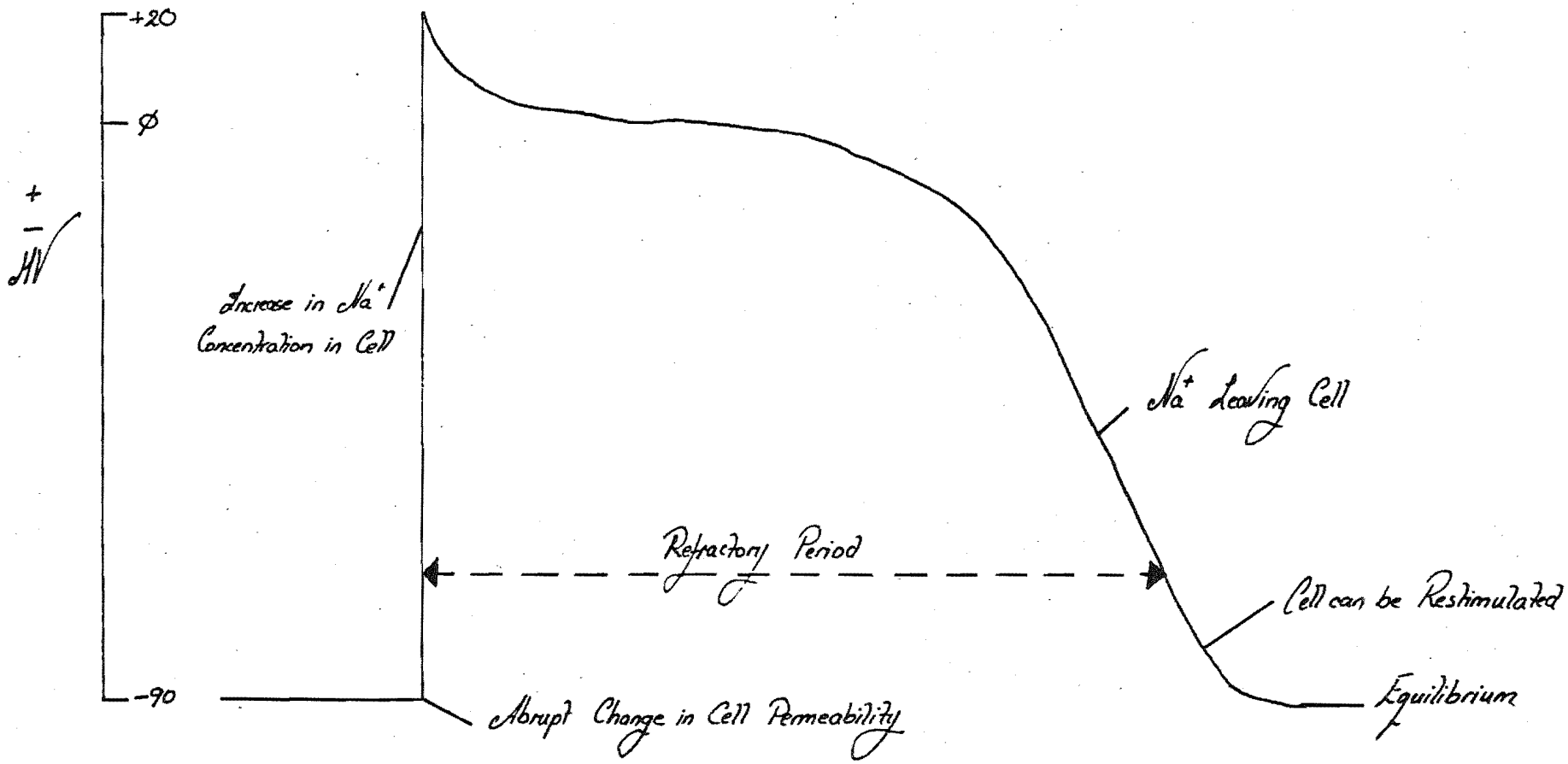
Through Muscle Cells

Fig C-3

adjacent cell that has just been stimulated. Thus there is a temporary blockage that travels immediatly behind the wave of stimulation (fig C-3). This blockage (called the refractory period) remains until the chemical pump can remove the excess of sodium ions within the cell and restore the relaxed cell's equilibrium. As this process continues the muscle cell is expanding and relaxing back to its original size.

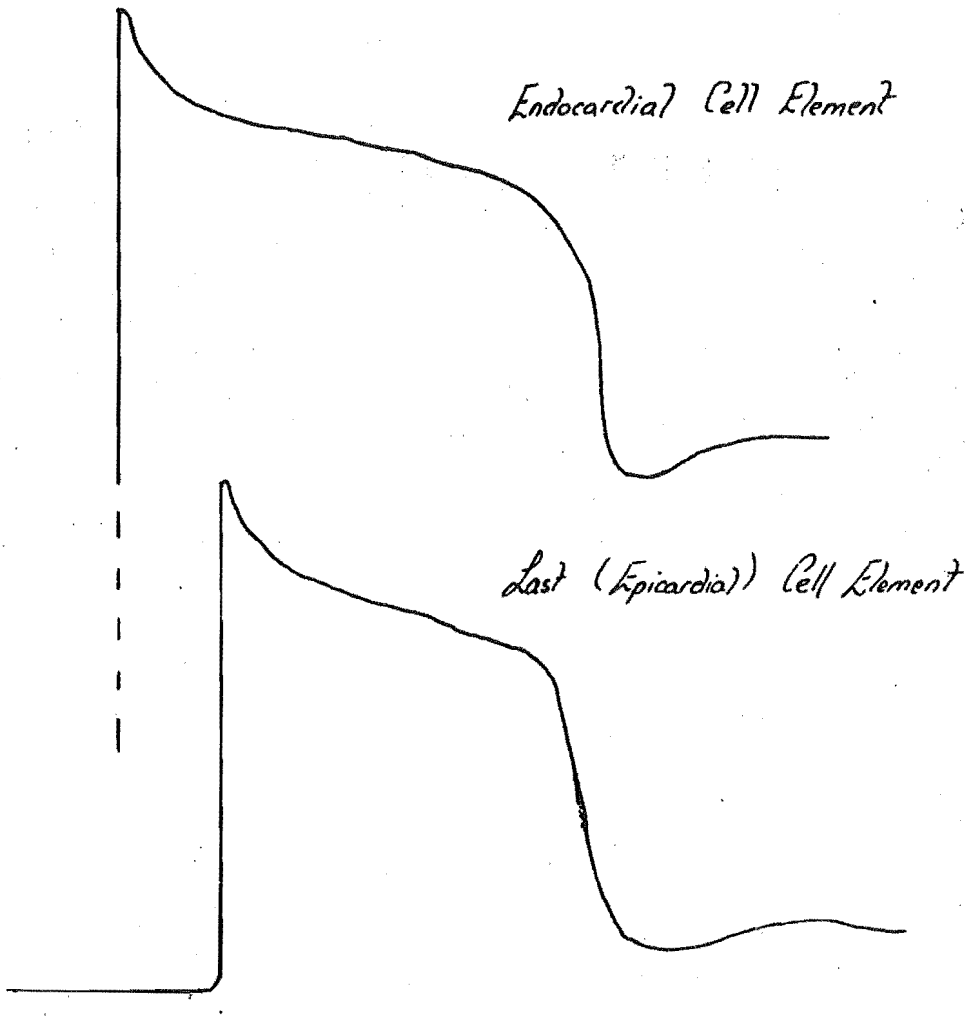
Figure C-4 shows the magnitude of the effective dipole source of a single muscle cell. The sum of the dipoles of all the cells within the muscle gives the gross electric field seen around the muscle. The wave form of a measurement of such a gross field is called an 'Electromyogram' except in the case of the heart muscle where it is called an 'Electrocardiogram'.

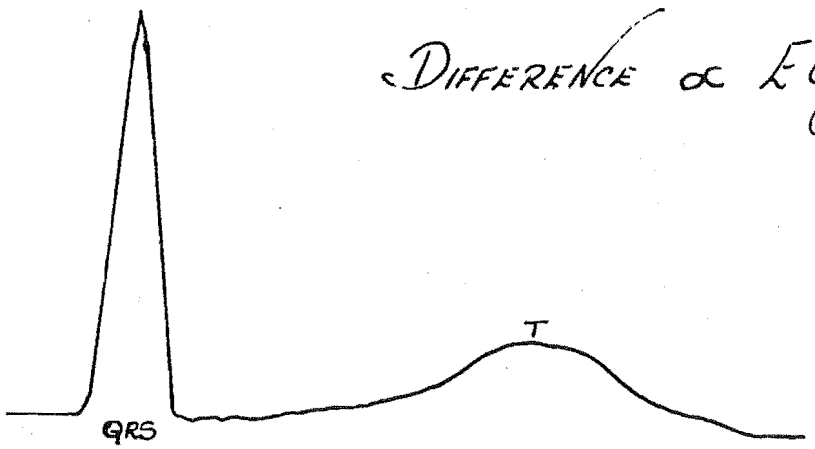
Podmanabhan [29] has shown how the ECG signal can be constructed from the wave forms of the muscle cells of the heart (fig. C-5).



Action Potential of Muscle Cell

Fig C-4





Derivation of ECG Signal

Fig C-5

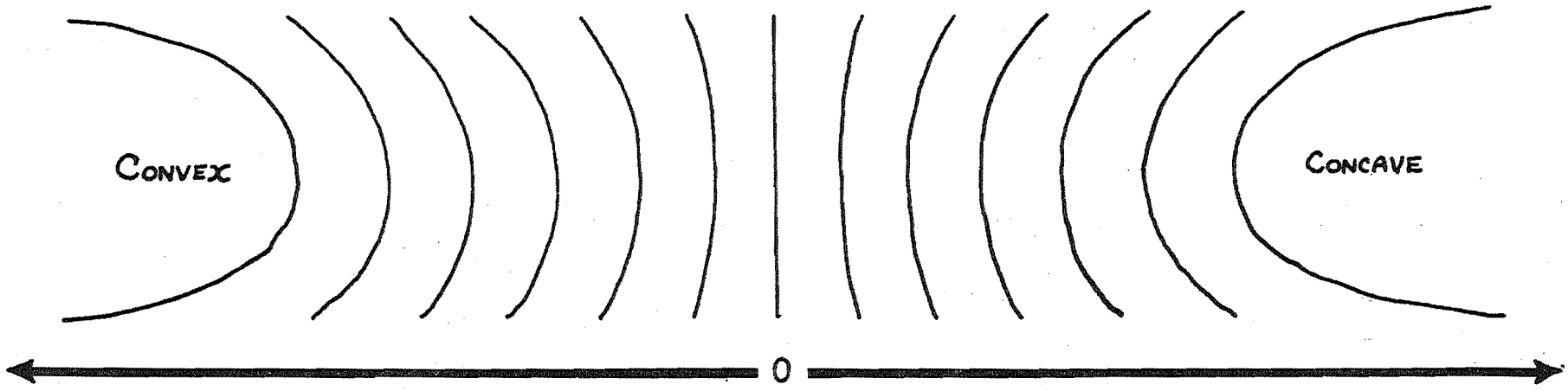
APPENDIX D

The Effect of the Threshold in Template Matching.

Section 3.3.2 introduced a threshold into the correlation matching scheme between a beat and the stored templates. The effect of the threshold is to allow slight differences between the wave shapes that are negligible for a particular analysis, but there is a potential problem that is more easily seen in the following analogy.

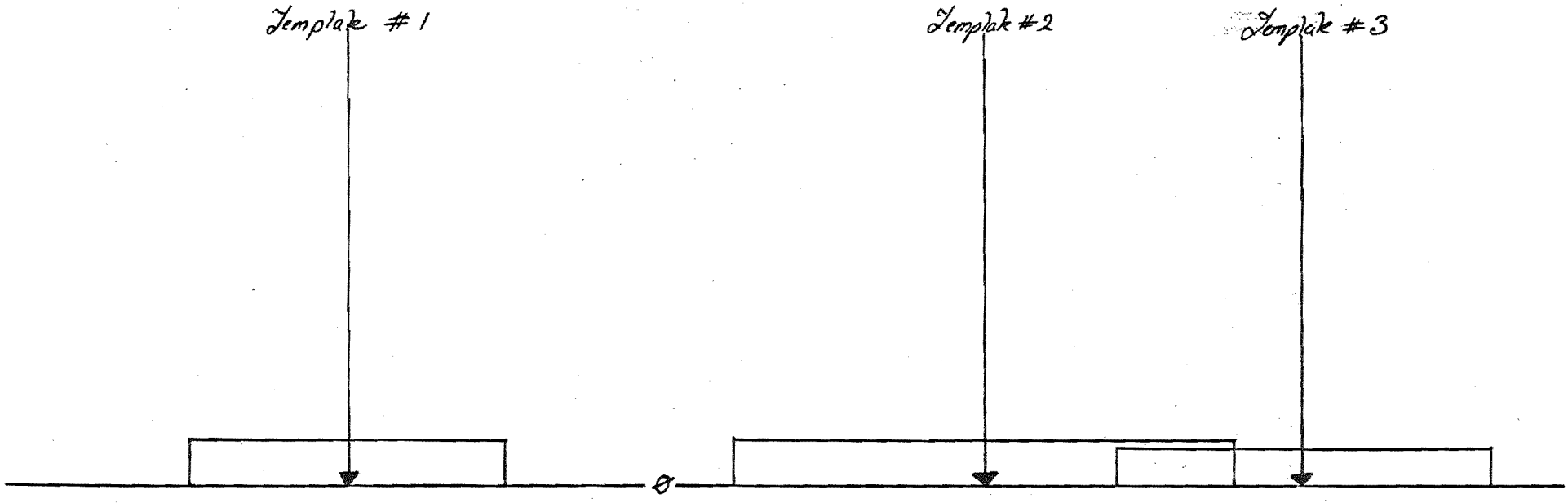
Consider a continuum that represents the amount of curvature of a line, with a straight line at the centre, convex lines to one side and concave lines to the other (fig D-1). If it is desired to group a selection of such curves using the algorithm outlined in this thesis then each template can be considered to be a point on the continuum and the value of its threshold can be seen to be a measure of the extent to which curves on either side of the template can be said to be similar (fig. D-2).

It is possible that the first curve of a new group that is encountered does not fall in the centre of an actual group, but is off to one side (fig. D-3). When a later curve from the other end of the actual group is found the operator could either extend the range of the existing template or create a new template. The first option



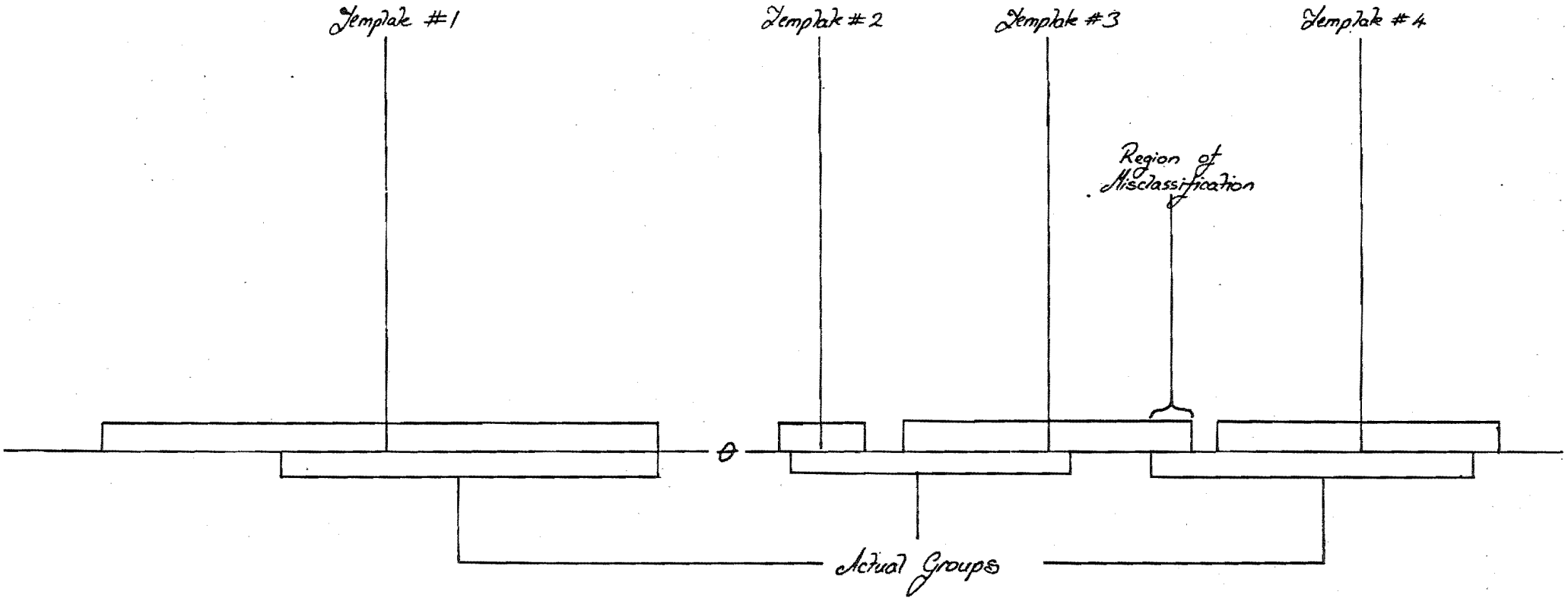
A CONTINUUM OF CURVED LINES

Fig D-1.



Effective Width of Template Due To Threshold

Fig D-2.



Difference Between Actual Groups and Groups Found.

Fig D-3.

introduces the possibility of the extended range of the template crossing into a neighbouring template' range while the second option means that two templates are used to represent one actual group. The latter option will usually be the better of the two as the two groups can be merged together at the end of the analysis.

The solution to this potential problem ultimatly lies in the ability and experience of the operator to judge the proper course of action given a particular set of templates.

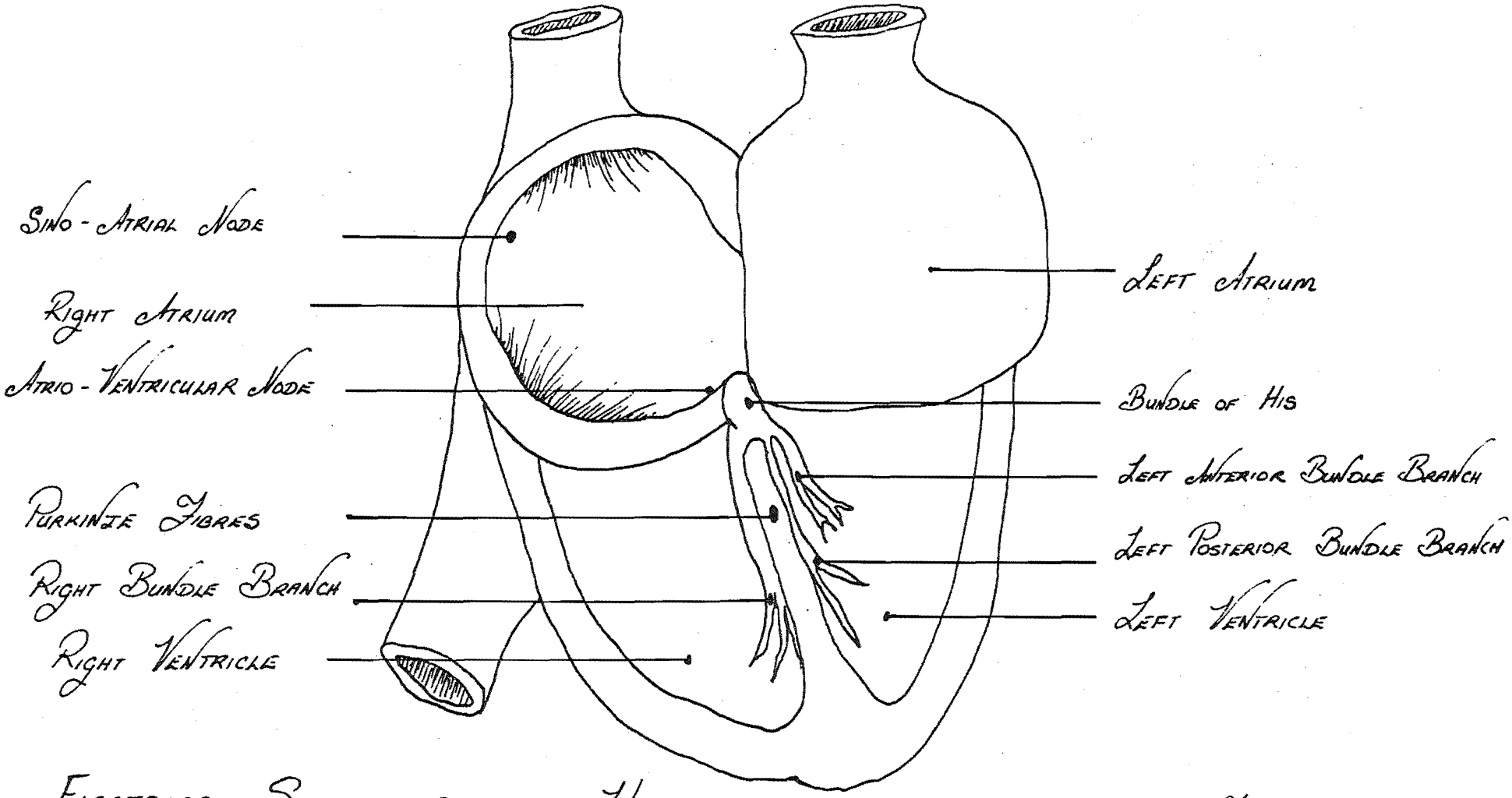
APPENDIX E

Function of The Heart.

Most people realise that the heart is one of the most important organs in the body in that it is required to pump the blood all around the body. Blood carries oxygen and nutriment to all the body's cells and carries away any waste products from the cells. It is therefore vital that the blood be kept moving. However, many people do not know the mechanism by which their hearts beat, or its relationship to the ECG tracing. Thus a short digression into physiology and electrophysiology will be made.

Physiological Description of the Heart

The human heart is separated into 4 distinct chambers, 2 on the left and 2 on the right (fig E-1). The two sides operate simultaneously and are similarly constructed. The right side takes de-oxygenated blood from the body and sends it to the lungs, while the left side takes the reoxygenated blood from the lungs and sends it back to all parts of the body.



Electrical Structure of the Heart

Fig E-1.

The upper chambers of the heart (the 'Atria') collect the incoming blood which is flowing at a low pressure and at a fairly constant rate. At the start of a heart beat the atria contract with a squeezing motion starting at the top of the heart and spreading in the general direction of the connecting valves through to the lower chambers (the 'Ventricles'). The ventricular muscle is relaxed, and expands to accomodate the blood from the atria. At the end of the atrial pumping action, the slightly pressurised ventricular blood tries to flow back into the atria but is prevented from doing so by the one-way valve through which the blood has just passed.

A short time later the ventricles start to contract from the lower regions of the heart and the blood is forced out of the heart to the lung and the rest of the body.

Electrophysiological Description of the Heart

Electrically the beat begins at the 'Sinoatrial node' (fig E-1), located at the top right hand corner of the heart, and spreads down through the atria towards the ventricles. However it is stopped from moving into the ventricular regions by a layer of fat tissue. In the middle of the fat layer is the 'Atrioventricular node' which delays the electrical stimulus for approximatly 200 mSec, before sending it down a specialised conduction system made from the 'His bundle', the 'Purkinje fibres' and the three major conduction bundles. The conduction bundles carry the stimulus to the whole lower heart region in such a way that the stimulus arrives at all parts within a small

fraction of time. A wave of contraction then spreads upward towards the outlet valves until it again strikes the fatty insulating layer just below the atria.

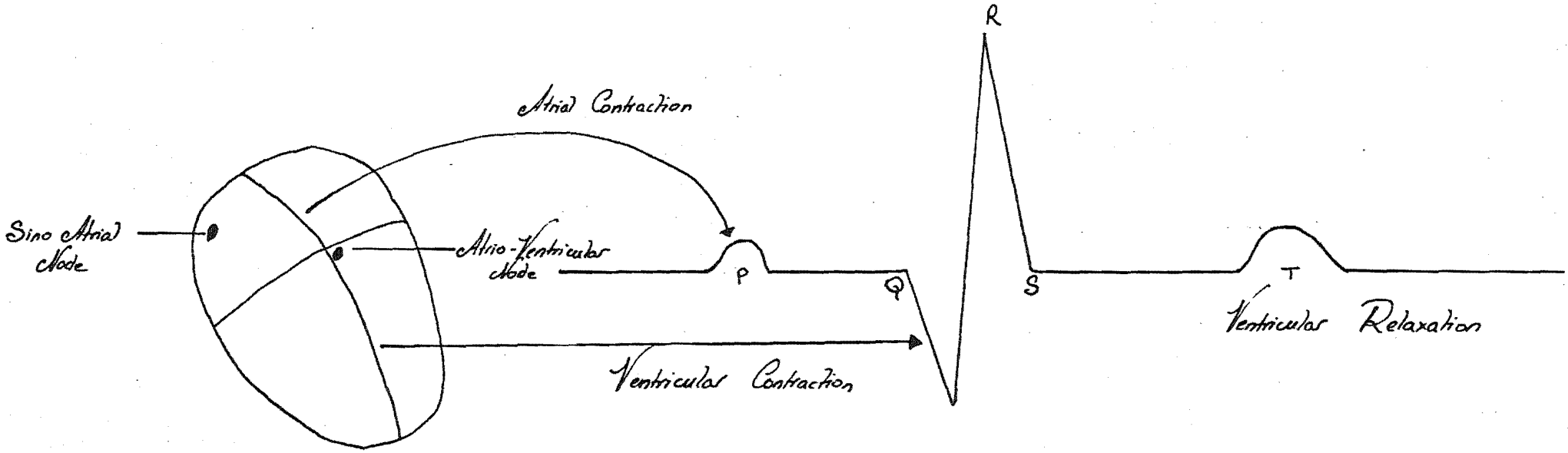
ECG Representation of a Heart Beat

Appendix C descibes how the ECG signal is generated by the muscle cells of the heart forming effective dipole sources while contracting and relaxing, and the vector sum of which is measured on the body surface.

Figure E-2 shows a classical ECG waveform and associates each feature with an event in the heart beat.

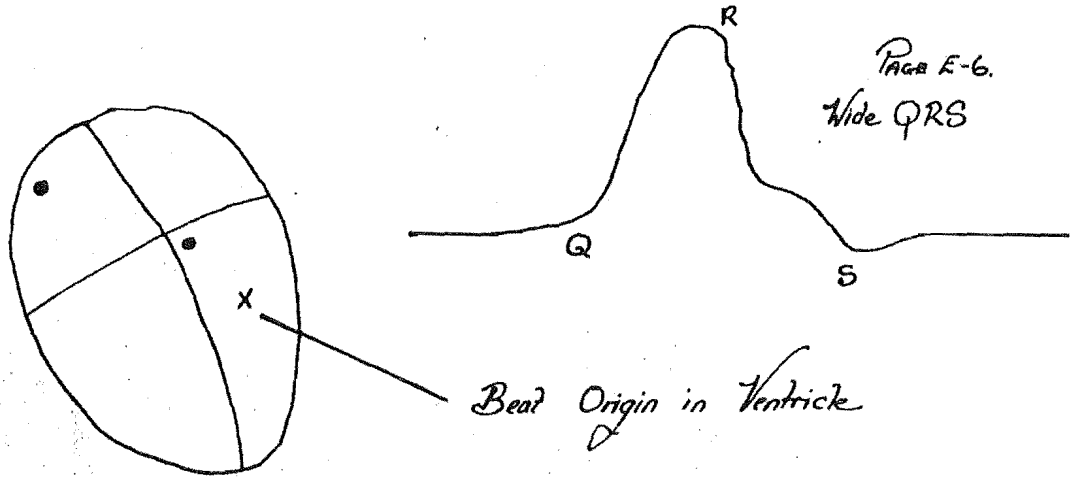
The ECG is a three dimensional vector moving within the body and so the placing of two electrodes on the body will give a projection of the vector onto a plane defined by the heart and the two electrodes. Thus the position of the electrodes on the body varies the shape of the ECG signal obtained and by careful choice of the electrode sites, a feature on the waveform can be accentuated or diminished.

If the heart beat does not follow the 'normal' path through the heart muscle then it is clear that the effective ECG vector will differ and so will the ECG trace. If the lead placement was favourable, an estimate of the actual abnormal conduction path through the heart can be made. Figure E-3 shows how some typical abnormalities can originate and how they are detected by the ECG

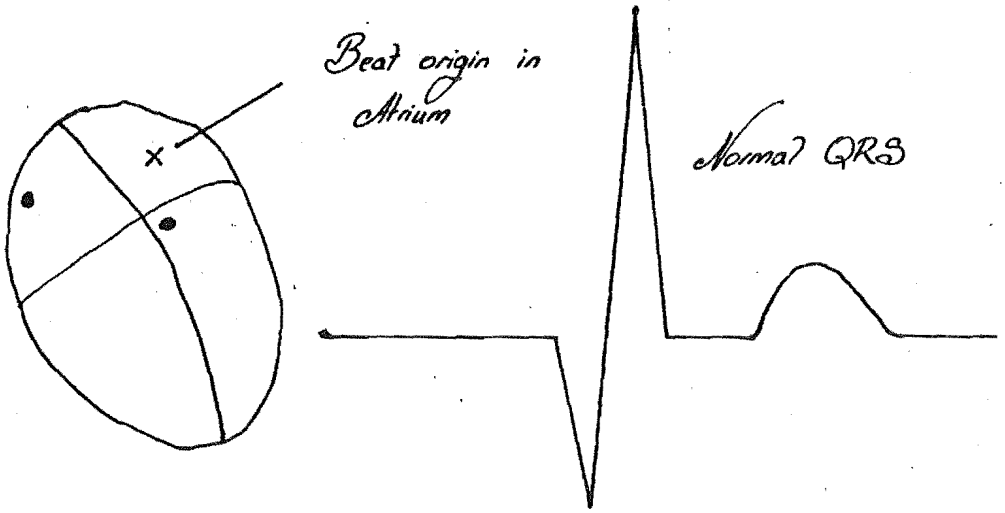


Relationship Between Heart Action and ECG Signal

Fig E-2



Ventricular Premature Beat



Atrial Premature Beat

Origins of Some Typical Abnormalities

Fig E-3

trace.

APPENDIX F

Results of a Trial ECG Analysis by R-R Interval Only.

Certain common amd important arrhythmias have characteristic R-R interval changes associated with them. The results presented here come from a program written to attempt to detect and classify these arrhythmias. As explained in section 3.2.1 this method was not developed beyond the initial stages due to the indeterminate nature of many of the results.

The measurement technique employed was to count the number of 10 mSec clock pulses received between successive QRS complexes to provide the basic R-R interval time. The tape replay hardware supplied a trigger pulse coincident with a detected QRS complex.

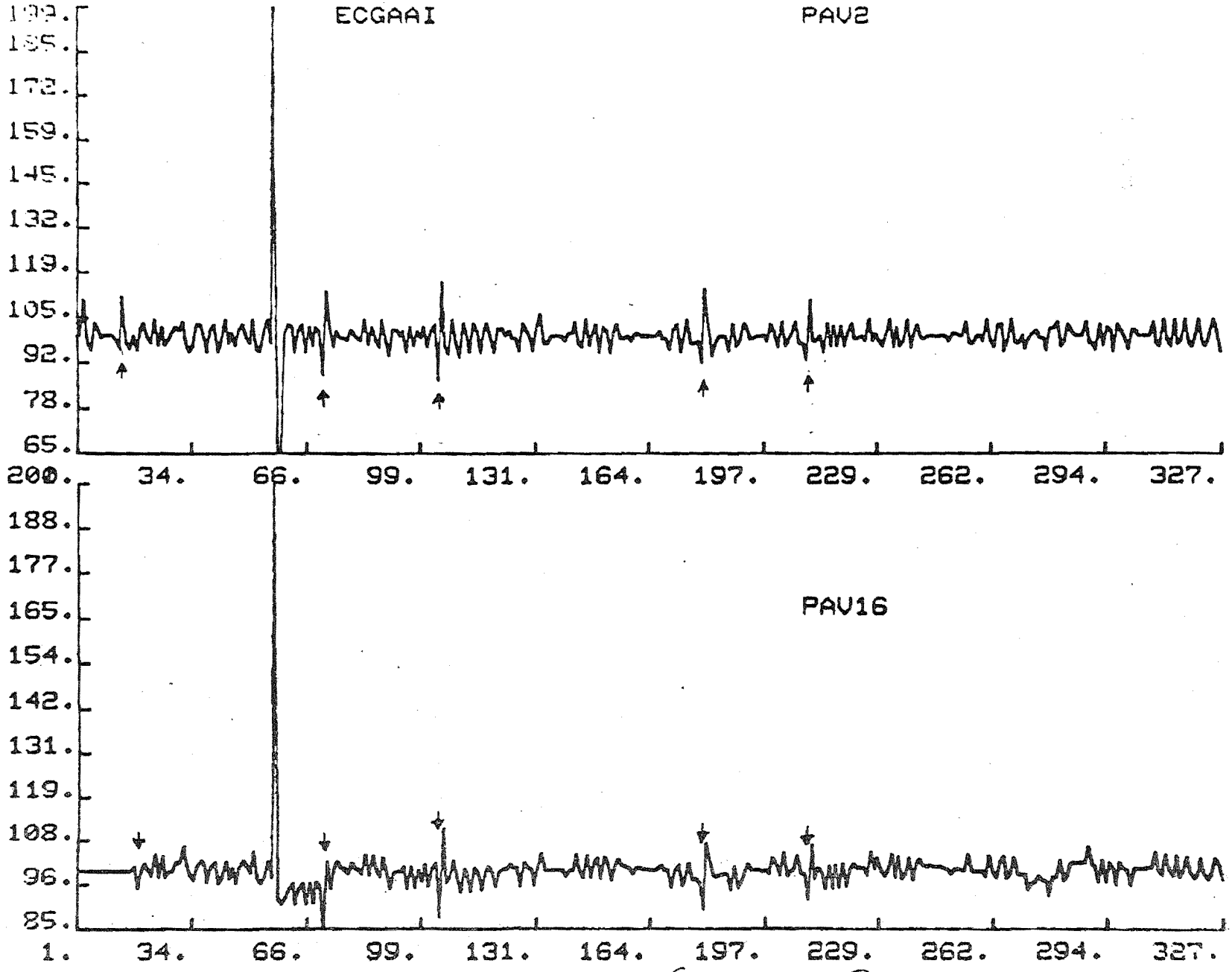
From the basic R-R interval eight derived measurements were calculated in order to determine which of the eight would be best suited for the arrhythmia detection. The first four of the derived values were the averages of the previous 2, 4, 8 and 16 intervals. The averages included all intervals even if some were previously detected as being abnormal.

The two beat average was used in an attempt to quickly follow any rate change due to normal physiological variations and yet to signal single abnormal R-R intervals. The sixteen beat average was expected to gloss over single abnormal beats but to indicate any rate changes with a rapid, ie 3 or 4 beat, onset. Such changes are generally caused by important abnormalities in the conduction path of the beat within the heart. The four and eight beat averages were calculated to determine if either of these gave better results than the two extreme averages.

The remaining four derived measurements were the current R-R interval expressed as a percentage of each of the calculated averages. The percentage values were used to determine when a particular event occured, independant of the absolute heart rate. Detection of an event was to be by the calculated percentage crossing either an upper or lower threshold.

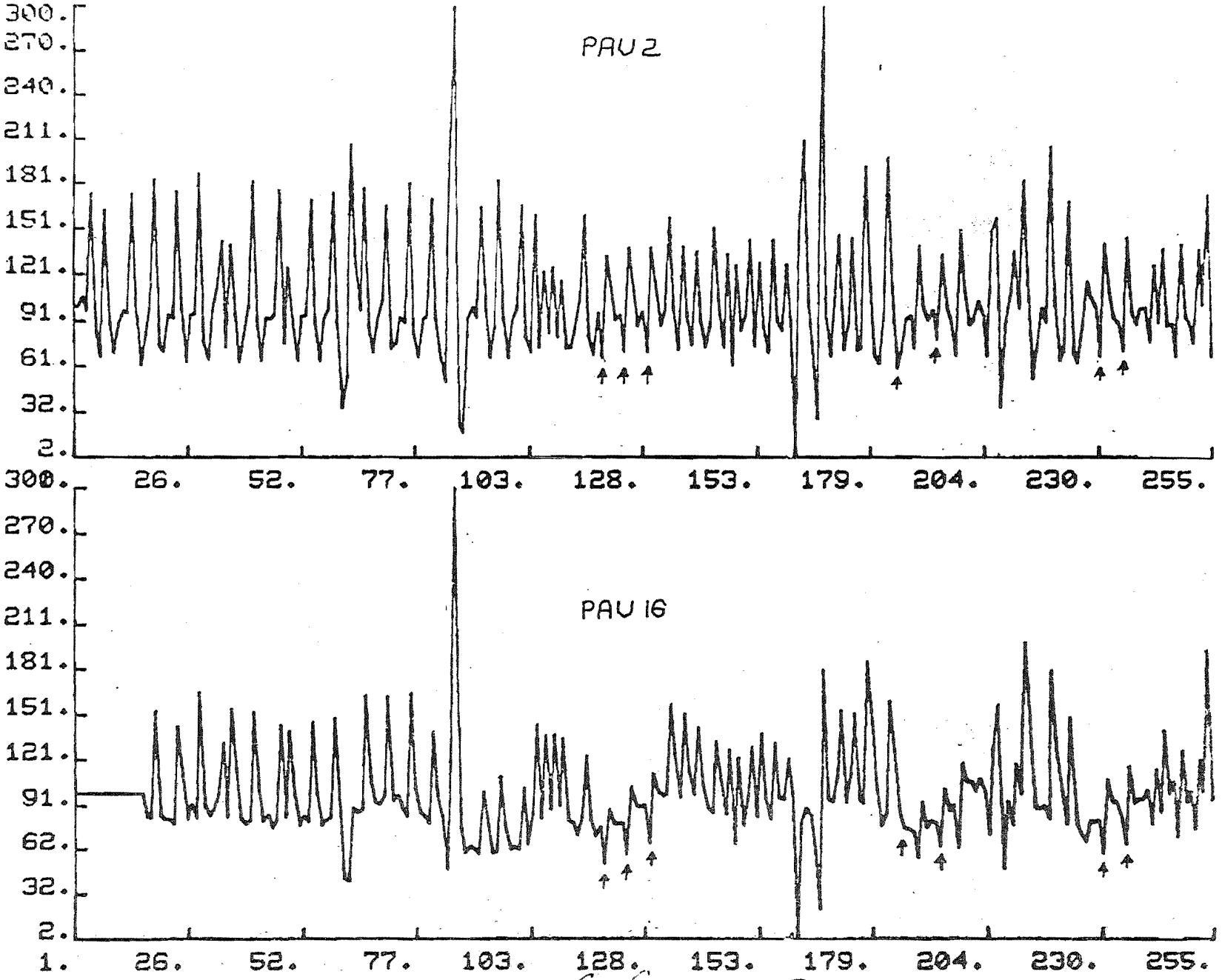
Figures F-1 and F-2 show plots of the percentage values derived from the 2 (PAV2) and the 16 (PAV16) beat averages for two tape segments. The plots are percentage R-R interval against detected QRS complex sequence number (NB time is not linear).

Figure F-1 shows how ventricular premature beats (VPB's) are detected. The characteristic of a VPB is an early beat followed by a compensating late beat so that the total period is two normal periods (see also table 3-1). The large spike at beat 57 is due to a complex that the hardware QRS detector failed to note.



R-R Interval Output Plot (1)

Fig 7-1.



R-R Interval Output Plot (2)

Fig 7-2

Figure F-2 shows how atrial premature beats (APB's) show up and also show the major problem with this analysis method. The APB's have the characteristic of an early beat that is followed by beats with normal periods (see also table 3-1). The unmarked peaks are due to a normal condition called sinus arrhythmia (see appendix A) and completely mask any attempt to differentiate the normal and abnormal characteristics.

APPENDIX G

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